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# The effect of heel height on frontal plane joint moments, impact acceleration, and shock attenuation during walking

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The effect of heel height on frontal plane joint moments, impact acceleration,  
and shock attenuation during walking

by

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A thesis submitted to the graduate faculty  
in partial fulfillment of the requirements for the degree of  
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## CHAPTER 1. GENERAL INTRODUCTION

### Introduction

Walking characteristics can be altered by footwear design properties, such as cushioning, stiffness, heel width and height. A commonly worn type of women's footwear that dramatically alters the kinematics and kinetics of walking is high-heeled shoes. Many women sacrifice the comfort and support of a typical shoe in order to sport the fashionable, aesthetically-pleasing stiletto heel. In fact, according to a survey done by the American Podiatric Medical Association, over 72% of women report wearing high heels, with 40% of these women wearing them daily (American Podiatric Medical Association, 2003, as cited in Yoon, An, Yoo, & Kwon, 2009).

Not only do high heels cause decreases in subjective comfort and support (Hong, Lee, Chen, Pei, & Wu, 2005; Yung-Hui & Wei-Hsien, 2005), but they alter many aspects of walking as well. Kinematically, increased heel height contributes to slower self-selected walking speeds and shorter stride lengths (Opila-Correia, 1990a; Esenyel, Katlen, Walden, & Gitter, 2003; Murray, Kory, & Sepic, 1970) and greater knee flexion, plantar flexion, trunk extension, and anterior pelvic tilt (Ebbeling, Hamill, & Crusemeyer, 1994; Opila-Correia, 1990b; Murray et al., 1970; Snow & Williams, 1994). In walking kinetics, greater peak vertical and anterior-posterior ground reaction forces throughout stance, medial forefoot pressures, peak knee extensor moments, peak external knee adduction moments, and lower peak plantar flexion moments have been shown with higher heel heights (Esenyel et al., 2003; Hong et al., 2005; Snow & Williams, 1994; Nyska, McCabe, Linge, & Klenerman, 1999; Kerrigan, Todd, & Riley, 1998; Kerrigan et al., 2005).



### Adverse Effects of Altered Gait Mechanics

Possible adverse effects associated with altered gait mechanics when wearing high heels have been identified and studied as far back as the mid-1900s (Joseph & Nightingale, 1956; Joseph, 1968; Adrian & Karpovich, 1965). Many of these studies have emphasized short term consequences for the foot. For example, higher forces in the medial forefoot have been reported while wearing high heels, which have been associated with foot disorders such as hallux valgus (Nyska et al., 1999; Corrigan, Moore, & Stephens, 1993; Snow, Williams, & Holmes, 1992). Other negative effects include lower back pain and leg pain (Jang & Kim, 1998, as cited in Lee, Jeong, & Freivalds, 2001; Hyun & Kim, 1997), and increased risk for ankle sprains (Nieto & Nahigian, 1975). Although there are evident short-term effects of wearing high heels, research investigating possible long-term effects is limited. The observed kinematic and kinetic alterations due to heel height may contribute adverse loading at the joints of the lower extremity, which may be an important factor in the development and progression of chronic conditions. Specifically, previous research has proposed wearing high heels contributes to joint degeneration and the development of knee osteoarthritis (OA) through repetitive dynamic loading during walking (Kerrigan et al., 1998; Kerrigan et al., 2005; Voloshin & Loy, 1994).

Knee OA affects 10% of adults over age 55 (Peat, McCarney, & Croft, 2001) and occurs in women more often than men (Dillon, Rasch, Gu, & Hirsch, 2006; Jordan et al., 2007). In addition, the prevalence of knee OA is increasing due to the aging population and higher rates of obesity (Lawrence et al., 2008; Coggon et al., 2001). Despite the obvious possibility of footwear contributing to joint degeneration, few studies have investigated the

relationship between wearing high heels and knee OA. Most studies examining the effect of heel height on walking mechanics have been limited to sagittal plane kinematics and kinetics.

#### Frontal Plane Kinetics

Although sagittal plane kinematic and kinetic alterations associated with high heel use have important implications for knee loading, many recent studies on joint degeneration have focused on the knee in the frontal plane and possible loading asymmetries within the joint that may contribute to knee OA. Knee OA occurs most often in the medial tibiofemoral compartment (Dearborn, Eakin, & Skinner, 1996; Keyes, Carr, Miller, & Goodfellow, 1992), and it's been suggested that greater medial loading on the knee may be a contributing factor in knee OA (Sharma et al., 1998). Schipplein and Andriacchi (1991) demonstrated that unequal forces are transmitted between the medial and lateral compartments during normal walking and reported medial forces may be as much as 2.5 times greater than those on the lateral side. Higher external knee adduction moments, also known as external knee varus moments, are representative of larger medial loads (Baliunas et al., 2002; Schipplein & Andriacchi, 1991) and are thought to be a significant component in the development of knee OA (Sharma et al., 1998; Baliunas et al., 2002; Schipplein & Andriacchi, 1991).

Previous research has reported higher external knee adduction moments in high-heeled walking compared to walking in lower heel heights (Kerrigan et al., 1998; Kerrigan et al., 2005; Esenyel et al., 2003). Kerrigan et al. (1998) found a significantly higher peak external varus moment (i.e. external adduction moment) during walking in 6 cm heels ( $0.32 \pm 0.07 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$ ) compared to barefoot walking ( $0.26 \pm 0.06 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$ ). In a follow-up study using wide-heeled, moderate height (3.8 cm) shoes, Kerrigan et al. (2005) found peak external knee varus moment during late stance was significantly higher for heeled shoes

than the same style control shoes of 0 cm heels. Peak varus moment during late stance was 14% higher for young women and 9% higher for elderly women. In addition, Esenyel et al. (2003) reported peak external knee adduction moment was 25% higher in a wide-heeled shoe of 6 cm heel height compared to a sport shoe with 1 cm heel height.

#### Impact Acceleration and Shock Attenuation

During normal walking, the body is subjected to an impact force at heel strike that is transmitted vertically through the ankle joint, tibia, and up through the skeleton in the form of a shockwave (Radin, Paul & Rose, 1972; Wakeling, Liphardt, & Nigg, 2003). This repetitive impulsive loading has been shown to be important in joint degeneration (Radin et al., 1972; Wosk & Voloshin, 1981; Collins & Whittle, 1989). The body attenuates the transient shock as it's transmitted up the body in order to minimize accelerations at the head, protecting the visual and vestibular functions needed in walking (Voloshin & Wosk, 1982; Ratcliffe & Holt, 1997; Menz, Lord, & Fitzpatrick, 2003). It has been suggested that the body's natural shock absorbers, which include soft tissue, bone, and joint structures such as menisci, articular cartilage, and intervertebral discs (Wosk & Voloshin, 1981; Collins & Whittle, 1989), progressively weaken due to this constant cyclical loading, allowing for initial damages in the knee joint and further joint degeneration due to compromised protection (Radin et al., 1972; Folman, Wosk, Voloshin, & Liberty, 1986; Voloshin, 1988). Natural shock absorbers weaken due to the gradual accumulation of microfractures in subchondral bone trabeculae, which lead to thickening and stiffening of the bone, decreasing its shock-absorbing capacity (Voloshin, 1988; Folman et al., 1986).

Footwear, when appropriately designed, can supplement the natural shock absorbers by contributing to further shock attenuation, as evidenced by reduced peak impact force and

reduced transient acceleration at the tibia (Lafortune & Hennig, 1992; Light, McLellan, & Klenerman, 1980). Further research on footwear properties has shown better shock attenuation in shoes with more compliant insoles (Light et al., 1980; Voloshin, 1988). Unlike most footwear, the design of high-heeled shoes limits their ability to attenuate shock during walking. For example, Voloshin and Loy (1994) reported higher tibial accelerations as heel height increased. Among three subjects, tibial accelerations ranged from 1.87 to 2.86 g in flat shoes of approximately 1.5 cm heel height to 3.45 to 4.51 g in heels approximately 7 cm in height (Voloshin & Loy, 1994). Though not specifically studied by Voloshin and Loy (1994), shock transmission can be quantified by measuring accelerations from multiple locations on the body. A common location that has been previously utilized to quantify shock attenuation throughout the body is the forehead (Wosk & Voloshin, 1981; Light et al., 1980). During walking, shock attenuation by the musculoskeletal system as measured by the peak tibial and head accelerations have been reported to be between 70 and 80% (Wosk & Voloshin, 1981; Light et al., 1980). Because the body acts to minimize head acceleration (Menz et al., 2003; Voloshin & Wosk, 1982; Ratcliffe & Holt, 1997), higher tibial accelerations imply a greater need for shock attenuation by the body's natural shock absorbers, which may contribute joint degeneration.

#### Limitations of Previous Research

Of the research on high-heeled gait, Kerrigan et al. (1998), Kerrigan et al. (2005), and Voloshin and Loy (1994) have most strongly emphasized the possible contributions to joint degeneration and knee OA. External varus moment (i.e. external adduction moment) of the knee was higher when walking in heels in comparison to barefoot walking (Kerrigan et al., 1998) and flat-heeled shod walking (Kerrigan et al., 2005). Although Kerrigan and

colleagues (1998; 2005) effectively identified possible characteristics of high-heeled gait that contribute to knee OA, these studies reflect several limitations. First, only two heel height conditions were employed in these studies, therefore the effect of systematically increasing heel height on peak external knee varus moment was not investigated. In addition, Kerrigan et al. (1998) used barefoot walking as their baseline condition. There are many inherent differences between barefoot and shod walking (Light et al., 1980; Morio, Lake, Guegen, Rao, & Baly, 2009), which raises questions about the value of contrasts between high-heeled and barefoot walking. Additionally, Kerrigan et al. (1998) provided little control over heel height and shoe design. In contrast, Kerrigan et al. (2005) controlled for shoe design by employing custom-made shoes only differing in heel height. They did not examine, however, any possible graded responses to multiple heel heights. Also, neither study controlled walking speed, allowing subjects to walk at preferred speeds for each heel height condition. Although a preferred walking velocity avoids imposing constraints on gait, enhancing real-world application, a fixed walking velocity controls for the speed effect on walking kinematics and kinetics.

Lastly, research investigating the effects of heel height on impact acceleration and shock attenuation throughout the body is limited. Voloshin and Loy (1994) reported results for a very limited analysis of three subjects. Although several heel heights in addition to a barefoot and flat (0.5cm) condition were investigated, heights varied among subjects. Thus, a true systematic manipulation of heel height was not included. In addition, shock attenuation, or the attenuation of the shockwave as it was transmitted up the body, was not measured by Voloshin and Loy (1994). Finally, both Kerrigan and colleagues (Kerrigan et al., 1998, 2005) and Voloshin and Loy (1994) only investigated heel heights up to

approximately 6 cm and 7 cm, which does not encompass the full range of heel heights typically worn by women.

### Purpose and Hypotheses

Links between wearing high heels and the development of degenerative joint disease have been proposed, specifically emphasizing a higher external knee adduction moment while wearing high heels and higher impulsive loading that occurs in the knee at heel strike. In addition, the effect of systematically increasing heel height on peak external knee adduction moment and impulsive loading at heel strike has not been investigated.

The purpose of this study was to determine the systematic effect of heel height on the net joint moments at the hip, knee, and ankle frontal plane during walking, with particular attention placed on the external knee moment. In addition, this study examined the effect of heel height on impact acceleration of tibia and head and shock attenuation of the body. It was hypothesized that frontal plane net joint moments of the lower extremity, specifically peak external knee adduction moment, increase systematically as heel height increases. It was also hypothesized that peak tibial acceleration increases systematically as heel height increases. Because of the need to maintain a stable head position during walking (Menz et al., 2003; Pozzo, Berthoz, & Lefort, 1990; Voloshin & Wosk, 1982; Ratcliffe & Holt, 1997), it was also hypothesized that shock attenuation increases systematically as heel height increases. As a result of substantial shock attenuation, it was predicted that peak head acceleration is unaffected by heel height.

## **Thesis Organization**

This thesis is organized into a general introduction chapter, two manuscript chapters, followed by a general conclusions chapter. Both manuscripts are formatted according to *Gait and Posture* specifications. The primary author for both articles is Danielle D. Barkema, a Master's student of Kinesiology at Iowa State University. Philip E. Martin is a co-author on both articles. Dr. Martin, Professor and Department Chair of Kinesiology at Iowa State University, assisted in research by contributing to the experimental design, data analysis, and preparation of the manuscripts. Dr. Timothy R. Derrick, Associate Professor at Iowa State University, is also a co-author on both articles. Dr. Derrick contributed to the experimental design, data processing and analysis, and preparation of the manuscript. Dr. Jason C. Gillette, Associate Professor at Iowa State University, and Dr. Rick L. Sharp, Professor at Iowa State University contributed to both articles in manuscript preparation.

## **Review of Literature**

### Introduction

Walking characteristics can be altered by footwear design properties, such as cushioning, stiffness, heel width and height. Typically, footwear has been designed to provide support for the foot and comfort for the wearer. However, in today's society comfort and support do not dictate all areas of shoe design. Appearance, fashion, and overall aesthetics sometimes take precedence over comfort in the design of many modern women's footwear, specifically high-heeled shoes. Over the past several decades, possible detrimental effects of wearing high heels have been identified and studied (Joseph & Nightingale, 1956; Joseph, 1968; Adrian & Karpovich, 1965; Nieto & Nahigian, 1975; Jang & Kim, 1998, as

cited in Lee et al., 2001), and yet the high heel shoe market has not been altered by these concerns. In fact, according to a survey done by the American Podiatric Medical Association (AMPA), over 72% of women report wearing high heels, with 40% wearing them daily (American Podiatric Medical Association, 2003, as cited in Yoon et al., 2009).

Wearing high heels has been shown to alter many walking characteristics, which is evident in the kinematics and kinetics of walking. Higher heel heights have been shown to increase knee flexion, plantar flexion, trunk extension, and anterior pelvic tilt (Opila-Correia, 1990b; Esenyel et al., 2003; Murray et al., 1970) in addition to result in slower self-selected walking speeds and shorter stride lengths (Opila-Correia, 1990a; Esenyel et al., 2003; Murray et al., 1970). Kinetically, greater peak vertical and anterior-posterior ground reaction forces throughout stance, medial forefoot pressures, peak knee extensor moments, peak external knee adduction moments, and lower plantar flexion moments have been observed during walking in high heels (Esenyel et al., 2003; Hong et al., 2005; Snow & Williams, 1994; Nyska et al., 1999; Kerrigan et al., 1998; Kerrigan et al., 2005). These observed alterations in kinematics and kinetics create potentially adverse loading at the joints of the lower extremity, which may be an important factor in the development and progression of chronic conditions (Kerrigan et al., 1998; Kerrigan et al., 2005; Voloshin & Loy, 1994; Nyska et al., 1999; Corrigan et al., 1993)

There have been proposed links to wearing high heels and lower extremity joint degeneration and the development and knee osteoarthritis (OA) through the loading at the knee frontal plane and through repetitive impulsive loading at heel strike (Kerrigan et al., 1998; Kerrigan et al., 2005; Voloshin & Loy, 1994). Specifically, external knee adduction moment and impact acceleration at heel strike are thought to be important components in



joint degeneration and knee OA (Sharma et al., 1998; Baliunas et al., 2002; Schipplein & Andriacchi, 1991; Radin et al., 1972; Wosk & Voloshin, 1981; Collins & Whittle, 1989). Higher external knee adduction moments and higher peak tibial accelerations at heel strike during walking have been shown with higher heel heights (Kerrigan et al., 1998; Kerrigan et al., 2005; Esenyel et al., 2003; Voloshin & Loy, 1994). It is important to recognize the effect heel height has on the kinematics and kinetics of walking in order to identify any potentially adverse effects that contribute to the development and progression of joint degeneration and knee OA.

### Knee Osteoarthritis

The degenerative disease of knee osteoarthritis is the most common form of osteoarthritis, affecting over 10 million Americans, with most being over the age of 45 (Parment, Lynn, & Glass, 2003). Women are more likely to suffer from knee OA than men (Dillon et al., 2006; Jordan et al., 2007), with estimates of 23.7 % of men and 31.0 % of women over age 45 suffering from knee OA (Jordan et al., 2007). Over the age of 60, approximately 31.2% of men and 42.1% of women suffer from knee OA (Dillon et al., 2006). Knee OA is also more common in ethnicities other than black or Hispanic (Jordan et al., 2007).

Knee OA is characterized by deterioration of the articular cartilage, which can be caused by previous knee injury, obesity, genetics, and repetitive strain on the knee. Acute knee injury, such as a ligament tear, fracture, or meniscal injury may predispose individuals to knee OA. Although risk estimates vary, a prospective study by Wilder, Hall, Barrett, and Lemrow (2002) found that women who had previously sustained an acute knee injury were 8.3 times as likely and men were 7.2 times as likely to develop knee OA compared to those

who did not have a history of knee injury. Similarly, obesity increases the risk for knee OA as well. Coggon et al. (2001) found that compared to a normal BMI of 24.0 to 24.9, the risk as defined by odds ratios of knee OA, progressively increased from 0.1 for a BMI less than 20, to 13.6 for a BMI greater than 36. If the history of knee injury was included, then the risk for knee OA increased to 13.1 for a BMI between 24.0 to 29.9 to 21.6 for a BMI greater than 30 (Coggon et al., 2001). In addition, Cooper et al. (2000) divided a sample of 99 men and 255 women into three BMI categories: less than 22.7, 22.7 to 25.4, and greater than 25.4. The odds ratio for knee OA using the lowest category as a reference group increased from 6.6 for the middle BMI group to 18.3 for the highest BMI group.

Genetics is also believed to play a role in the development and progression of knee OA. It is difficult to determine the heritability of knee OA, but there are estimates. Spector, Cicuttini, Baker, Loughlin, and Hart (1996) investigated radiographic OA in the knees of female twins and found the influence of genetic factor to be 39%. However, Bijkerk et al. (1999) estimated that heritability of knee OA was not significant at 7%. Finally, Zhai, Hart, Kato, MacGregor, and Spector (2007) reported the heritability of the progression of knee OA as determined by the presence of osteophyte and joint space narrowing using radiography to be 71% for monozygotic twins and 48% for dizygotic twins. The estimated magnitude of genetic influence in knee OA varies, but there is evidence of genetic factors contributing to the disease.

Frontal and sagittal plane knee kinetics. The final cause of knee OA, which is the factor present in all cases of knee OA, is repetitive knee loading. The knee joint is made up of the interactions of the femur, tibia, and patella, which combine to form three knee compartments: the medial and lateral tibiofemoral compartments, and the patellofemoral

compartment. Knee osteoarthritis may occur in any of these three compartments, but the medial compartment is the most common site (Dearborn et al., 1996; Keys et al., 1992). However, many people suffering from knee OA may experience it in multiple compartments, especially in severe cases (Ledingham, Regan, Jones, & Doherty, 1993).

There are several factors that may contribute to the high incidence of medial compartment knee OA. The medial compartment is characterized by thinner articular cartilage than the lateral compartment (Cicuttini et al., 2002), predisposing the compartment to further degeneration (Lewek, Rudolph, & Mackler-Snyder, 2004). In addition, Riegger-Krugh, Gerhart, Powers, and Hayes (1998) suggest that the lateral meniscus serves better to protect the lateral compartment in comparison to the medial meniscus on its compartment. Mechanically, medial compartment knee OA may be the most common due to exaggerated loads during walking. Walking generates unequal forces across the knee, with loads being up to 2.5 times greater on the medial compartment than the lateral compartment (Schipplein & Andriacchi, 1991). Greater loads on the medial compartment imply greater compression on the medial side of the knee, causing further degeneration of the joint (Sharma et al., 1998). In addition, greater loads at the medial compartment are associated with higher knee external adduction moments (Baliunas et al., 2002; Schipplein & Andriacchi, 1991). Subjects with medial compartment knee OA have been shown to walk with higher knee external adduction moments (Baliunas et al., 2002; Sharma et al., 1998). Sharma et al. (1998) showed that external knee adduction moment is strongly associated ( $r = 0.68$ ) with the progression of medial compartment knee OA based upon the Kellgren-Lawrence (K-L) radiographic grade.

Patellofemoral compartment knee OA often accompanies medial compartment knee OA, which is evident in Ledingham et al. (1993) finding that 50% of subjects with medial

compartment knee OA were also diagnosed with patellofemoral compartment knee OA. Patellofemoral compartment knee OA is often characterized by a malaligned knee cap, which often results from medial compartment knee OA (Cooke, Scuadmore, & Greer, 2000). Kerrigan et al. (2005) relates a prolonged external knee flexor moment from early to midstance to increased strain and pressure through the patella tendon and at the patellofemoral joint. Specifically, this occurs due to the work of the quadriceps during walking, leading to the conclusion that walking with increased quadriceps work may contribute to patellofemoral compartment knee OA (Kerrigan et al., 2005). Although forces in the sagittal plane are important in the development and progression of knee OA, recent studies on knee OA have focused on the frontal plane due to the high incidence of medial compartment knee OA.

#### Footwear and Frontal Plane Kinetics

In the frontal plane, it is evident that higher external knee adduction moments are key contributors to the greater loads on the medial compartment of the knee during walking (Sharma et al., 1998; Baliunas et al., 2002). Therefore, any alterations in the kinematics and kinetics of walking that result in these forces deserve attention. Because footwear can alter the kinematics and kinetics of walking, efforts have been made to reduce the joint loading that contributes to knee OA. Several footwear interventions have been tested in order to reduce the medial loading at the knee and have been shown to be successful (Fisher, Dyrby, Mundermann, Morag, & Andriacchi, 2007; Shakoor, Lidtke, Sengupta, Fogg, & Block, 2008). Fisher et al. (2007) found that increases in the lateral stiffness and the lateral thickness of tennis shoes, results in significantly lower peak external knee adduction moments in comparison to placebo walking shoes. Similarly, Erhart, Mundermann,

Mundermann, & Andriacchi (2008) reported a systematic increase in lateral wedge degrees led to a systematic decrease in peak external knee adduction moment. If shoe design interventions can reduce the forces present during walking at the knee, then it is logical to infer that certain types of footwear, specifically those not designed for comfort and support such as high heels, can enhance the forces at the knee, creating adverse effects.

Although research investigating heel height and possible links to joint degeneration is limited, Kerrigan et al. (1998) and Kerrigan et al. (2005) have emphasized the notion that altered kinematics and kinetics of walking in high heels contribute to knee OA. Kerrigan et al. (1998) showed a 23% higher external knee varus moment (i.e. external knee adduction moment) when walking in high heels with an average height of 6 cm compared to barefoot walking. Kerrigan et al. (2005) found that wide-heeled shoes with a moderate height of 3.8 cm resulted in significantly higher external knee varus moment by 14% in young women and 9% in elderly women compared to control shoes. A follow-up study to Kerrigan et al. (1998) investigated the effects of heel width in shoes averaging 7 cm in heel height on external knee varus moment and found there to be no significant difference between wide-heeled and narrow-heeled shoes (Kerrigan, Lelas, & Karvosky, 2001). External knee varus moment was 26% higher and 22% higher for the wide-heeled and narrow-heeled shoes, respectively. Additionally, Esenyel et al. (2003) reported the external knee adduction moment was 25% higher in a wide-heeled shoe of 6 cm heel height compared to a sport shoe with 1 cm heel height. In the sagittal plane, Kerrigan et al. (1998) and Kerrigan et al. (2005) reported a prolonged external knee flexor moment, which the authors deemed significant in the development of patellofemoral knee OA.

Impact acceleration and shock transmission. It is evident that repetitive knee loading may lead to unequal forces between the medial and lateral aspects of the knee, which are important in the development of medial compartment knee OA. Similarly, the repetitive axial loading that occurs at heel strike during walking has been shown to be significant in joint degeneration and the development of knee OA (Radin et al., 1972; Wosk & Voloshin, 1981; Collins & Whittle, 1989). At heel strike during normal walking, an impact force is transmitted through the lower extremity segments and joints and up the rest of the body in the form of a shockwave. Shockwave transmission creates peaks in the forces across the articular surfaces of the lower extremity joints and spine (Chu, Yazdani-Ardakani, Gradisar, & Askew, 1986). Because this force occurs over a short period of time (5 to 25 ms) (Folman et al., 1986), it is often referred to as an ‘impulsive load’ or ‘heel strike transient’ (Whittle, 1999). This repetitive impulsive loading that occurs during walking is believed to be a key mechanical factor in the development of knee OA (Radin et al., 1972; Wosk & Voloshin, 1981). The body attenuates the transient shock as it’s transmitted up the body in order to minimize accelerations at the head, protecting the visual and vestibular functions needed in walking (Voloshin & Wosk, 1982; Ratcliffe & Holt, 1997; Menz et al., 2003). The body’s natural shock absorbers, which include soft tissue, bone, and joint components, such as the meniscus, articular cartilage, and intervertebral discs (Wosk & Voloshin, 1981; Collins & Whittle, 1989), attenuate and dissipate the shockwave as it travels through the musculoskeletal system (Wosk & Voloshin, 1981). However, it has been suggested that the body’s natural shock absorbers progressively weaken due to repetitive impulsive loading, allowing for initial damages in the knee joint and further joint degeneration due to compromised protection (Radin et al., 1972; Folman et al., 1986; Voloshin, 1988).

Repetitive impulsive loading leads to the gradual accumulation of microfractures in the subchondral bone trabeculae (Radin et al., 1972). As part of the healing process, bone remodeling causes thickening and stiffening of the subchondral bone, decreasing its shock-absorbing capacity (Radin et al., 1972; Voloshin, 1988). The relatively unprotected overlying articular cartilage is subjected to increased stress from repetitive loading, leading to cartilage breakdown and joint degeneration (Radin et al., 1972; Voloshin, 1988; Folman et al., 1986).

Although no specific prospective studies involving humans have been documented, animal studies have provided evidence for this theory of joint degeneration. Guinea pigs and rabbits that were subjected to repetitive longitudinal impact loads developed degenerative joint disease in the knee after several weeks (Simon, Radin, Paul, & Rose, 1972; Radin et al., 1978). Radin, Orr, Kelman, Paul, and Rose (1982) showed that sheep that were walked on concrete floors for 30 months exhibited subchondral bone stiffening and eventually developed degenerative joint disease, whereas sheep walking on a soft surface did not.

Previous research has sought to quantify the transient force at heel strike and shockwave passing through the musculoskeletal system with the use of ground reaction force (GRF) measures and accelerometry. Impact force and acceleration measurements vary among previous studies for walking and running activities. During walking, peak impact forces have been shown to range from 0.44 BW up to 1.5 BW (Lafortune & Hennig, 1992; Wosk & Voloshin, 1981) and can rise up to 5 BW during running (Hreljac, 2004). In addition, peak tibial accelerations range from 1.71 to 3.6 g (Lafortune, 1991; Gill & O'Conner, 2003) during walking and up to 11 g during running (Perry & Lafortune, 1995). Shock attenuation by the musculoskeletal system, as measured by the peak tibial and head

accelerations, ranges from approximately 70 to 80% (Wosk & Voloshin, 1981; Light et al., 1980). Ratcliffe and Holt (1997) measured ankle and head accelerations, reporting shock attenuation of 92%. It is evident that reported peak impact forces, accelerations, and shock attenuation values differ among previous studies, which may be attributed to a variety of factors. However, an important component that has received a lot of attention in previous literature is the effect of footwear.

#### Footwear and Impact Acceleration and Shock Attenuation.

Footwear design properties can affect the loading at the lower extremity joints during walking. It is evident that repetitive impulsive loading and shock transmission during walking can result in joint degeneration, and efforts have been to reduce the loading on the lower extremity through footwear design. Compared to barefoot walking, shod walking has shown reduced peak impact force and peak tibial acceleration. For example, Lafortune and Hennig (1992) found that peak impact force was  $0.69 \pm .10$  BW and peak tibial acceleration was  $4.68 \pm 1.07$  g during barefoot walking compared to  $0.48 \pm .12$  BW and  $2.92 \pm .31$  g during walking in leather street shoes. A decrease in peak tibial acceleration from  $3.30 \pm .20$  g for barefoot walking to  $1.71 \pm .09$  g for shod walking was reported by Lafortune (1991) as well. Further research on footwear properties has shown better shock absorption of shoes with more compliant insoles (Lafortune & Hennig, 1992; Light et al., 1980; Voloshin, 1988). Walking in athletic shoes resulted in a peak tibial acceleration of  $2.52 \pm .26$  g compared to  $2.92 \pm .31$  g during walking in leather street shoes (Lafortune & Hennig, 1992). In addition, Voloshin (1988) found that the use of viscoelastic insoles in shoes during walking increased the shock absorbing capacity of the footwear up to 30%. Thus, footwear can act to supplement the body's natural shock absorbers by contributing to further shock attenuation.



The design of high-heeled shoes limits its ability to attenuate shock during walking. The rigid structure and lack of cushioning capability of most high-heeled shoes allow for less shock absorption at heel strike impact compared to the cushioned sole and insole of most footwear designs (Voloshin & Loy, 1994). Research addressing this topic is limited, but Voloshin and Loy (1994) found higher tibial accelerations during walking as heel heights increased. Tibial accelerations for heel heights of approximately 7.5 cm were 75% higher than shoes of approximately 1.5 cm heel height. Although not reported, heel heights of about 4 cm also resulted in higher peak tibial accelerations than those exhibited in flat shoes, producing a systematic trend for each subject as heel height increased. It was also noted that higher heel heights resulted in a more pronounced metatarsal strike that occurred approximately 50 ms after heel strike, which is defined as the metatarsal portion of the foot contacting the floor (Voloshin & Loy, 1994). For example, in high-heeled shoes of approximately 7 cm, peak tibial acceleration at heel strike was 3.45 g, which was followed by a tibial acceleration of 2.45 g at metatarsal strike. Therefore, high-heeled walking not only produces higher tibial accelerations at heel strike, but may result in a metatarsal impact that leads to increased loading at the lower extremity joints as well. In addition, increased peak impact force measurements have been reported with increasing height during high-heeled walking (Yung-Hui & Wei-Hsien, 2005; Hong et al., 2005; Stefanyshyn, Nigg, Fisher, O'Flynn, & Liu, 2000). Yung-Hui and Wei-Hsien (2005) found peak impact force to increase from  $0.46 \pm .15$  in flat shoes of 1 cm heel height to  $0.61 \pm .17$  BW in high-heeled shoes of 7.6 cm heel height. Despite most footwear contributing to shock attenuation during walking, the design high-heeled shoes allows for greater shock transmission to the lower extremity, which has shown to be significant in joint degeneration and knee OA.

### Heel Height Conditions

Previous research on heel height and walking has utilized several heel height conditions that vary in the range of heel heights and type of footwear used. In addition to many shoe heel heights employed, previous studies investigating heel height had subjects walk barefoot as a baseline condition (Kerrigan et al., 1998; Eisenhardt, Cook, Pregler, & Foehl, 1996; Corrigan et al., 1993; de Lateur, Giaconi, Questad, Ko, & Lehmann, 1991; Gefen, Megido-Ravid, Itzhak, & Arcan, 2002). Barefoot walking serves as convenient, plausible condition, but contrasts between shoe heel heights may not be of great significance as there are many inherent differences in the kinematics and kinetics between barefoot and shod walking (Morio et al., 2009; Wolf et al., 2008). For this reason, other heights have been utilized for baseline conditions. Flat shoes ranging from 0 to 1 cm heel heights served as baseline conditions for Yung-Hui & Wei-Hsein (2005), Kerrigan et al. (2005), Lee et al. (2001), Hong et al. (2005), Hansen and Childress (2004), Esenyel et al. (2003), and Ebbeling et al. (1994). These studies may have better controlled for the influence of shoes in general on the kinetics and kinematics of walking, but it is still difficult to select a heel height representative of a commonly worn flat shoe. Other studies have not included a flat heel condition and investigated varying heights of high heels (Eisenhardt et al., 1996; Nyska et al., 1996; Speksnijder, Munckhof, Moonen, & Walkenkamp, 2005; Opila-Correia, 1990a; Opila-Correia, 1990b; Stefanyshyn et al., 2000; Snow & Williams, 1994; Mandato & Nester, 1999).

Ranges of heel heights vary among previous studies. According to the American Podiatric Medical Association, heels of two inches (5.08 cm) are considered “high heels” (American Podiatric Medical Association, 2003, as cited in Yoon et al., 2009), although there are differing opinions about what constitutes a “high heel.” Due to the possible interaction

between a systematic increase of heel height and a graded response in variables, Yung-Hui and Wei-Hsien (2005), Hong et al. (2005), Lee et al. (2001), and Snow and Williams (1994) used three heights ranging from 1 to approximately 8 cm. In addition, Eisenhardt et al. (1996), Voloshin and Loy (1994), Stefanyshyn et al. (2000), and Ebbeling et al. (1994) used four heights ranging from just over 1 cm to about 8 cm, with Eisenhardt et al. (1996) and Voloshin and Loy (1994) employing a barefoot baseline condition. Corrigan et al. (1993) selected lower heights of 2 and 4 cm with a barefoot baseline condition. Other studies have investigated the difference between low and high heels using 1.95 and 5.91 cm (Speksnijder et al., 2005), 1.6 and 6.1 cm (Opila-Correia, 1990a; 1990b), and 5 and 8 cm (Mandato & Nester, 1999). Kerrigan et al. (2005) and Kerrigan et al. (1998) studied two heights, using a control against a shoe of 3.8 cm height, and a barefoot baseline condition against a 6-cm heel height, respectively. The effect of the systematic manipulation of heel height resulted in several variables showing a graded response in Snow and Williams (1994), Eisenhardt et al. (1996), Stefanyshyn et al., (2000), Ebbeling et al. (1994), Lee et al. (2001), and Corrigan et al. (1993). The use of a baseline condition and a selected heel height may provide insight into the effect of heel height on walking, but the use of multiple heights may be more beneficial in order to observe the effects of systematically increasing heel height.

A variety of types of footwear have been used to study the effect of heel height on walking. Most studies used high heels with narrow toe boxes and narrow heels and of similar construction for each height (Ebbeling et al., 1994; Stefanyshyn et al., 2000; Hong et al., 2005, Yung-Hui & Wei-Hsien; Lee et al., 2001; Snow & Williams, 1994). Opila-Correia (1990a; 1990b) and Speksnijder et al. (2005) did not control for design by instructing participants to wear their own heels, averaging the heights in order to study the effect of heel

height. Kerrigan et al. (1998) and Nyska et al. (1996) instructed subjects to wear their own footwear but only included designs with a narrow heel. Other studies have employed different styles of heels as evident in Esenyel et al. (2003) using a 5.5 cm wide-heeled dress shoe, Kerrigan et al. (2005) using a custom-made control and wide-heeled dress shoe, and de Lateur et al. (1991) using a heel-strapped clog shoe. Esenyel et al. (2003) and Mandato and Nester (1999) used a sport shoe for comparison, despite the obvious lack of similarity in construct with most high-heeled shoes. Lastly, Corrigan et al. (1993) did not use shoes and instead attached rigid polyurethane heels by bandaging them to the foot, limiting the real-world application because actual shoes were not worn, but providing much better control over heel design.

Overall, previous research has employed a variety of ranges of heel heights and types of footwear. In order to avoid confounding variables, baseline condition of 0 to 1 cm and of similar construct to the heels being studied may be of benefit for future studies. In addition, the use of multiple heights in order to study the effect of systematically increasing heel height may be of interest.

### Speed

There are conflicting results on the effect of heel height on walking speed and on the practice of controlling for speed. Opila-Correia (1990a; 1990b), and Esenyel et al. (2003) found reductions in walking speed ranging from 5.4 to 7.5% with high heels compared to controls. However, Kerrigan et al. (2005) did not find any significant differences in walking speed when comparing 0 cm shoes compared to 3.8 cm heels. Because of the possible effect of heel height on walking speed, studies have differed in controlling for walking. Walking speed influences the kinematics and kinetics of gait (Stoquart, Detrembleur, & Lejeune,

2008; Lelas, Merriman, Riley, & Kerrigan, 2003), supporting the practice of controlling speed. Ebbeling et al. (1994) controlled for speed at  $4.2 \text{ km}\cdot\text{h}^{-1}$  ( $1.17 \text{ m}\cdot\text{s}^{-1}$ ), while Lee et al. (2001) and Nyska et al. (1996) used  $4 \text{ km}\cdot\text{h}^{-1}$  ( $1.11 \text{ m}\cdot\text{s}^{-1}$ ). In addition, Yung-Hui and Wei-Hsien (2005) and Hong et al. (2005) used  $1.3 \text{ m}\cdot\text{s}^{-1}$  and Snow and Williams (1994) and Stefanyshyn et al. (2000) used  $1.4 \text{ m}\cdot\text{s}^{-1}$ . Eisenhardt et al. (1996) used a cadence of 100 steps/min, Voloshin and Loy (1994) used a cadence of approximately one walking cycle per second, and Hansen and Childress (2004) had subjects walk at preferred, “slow” and “fast” speeds without reporting averages. In contrast, the use of preferred walking speed enhances real-world application by avoid imposing constraints on gait, therefore, many previous studies have allowed self-selected walking speeds (Corrigan et al., 1993; de Lateur et al., 1991; Esenyel et al., 2003; Kerrigan et al., 1998; Kerrigan et al., 2005; Mandato & Nester, 1999; Speksnijder et al., 2005, Opila-Correia, 1990a; 1990b). Opila-Correia (1990a) and Esenyel et al. (2003) provided average speeds and amongst conditions  $1.37 \pm .13 \text{ m}\cdot\text{s}^{-1}$  for low heels and  $1.28 \pm .12 \text{ m}\cdot\text{s}^{-1}$  for high heels, and  $1.29 \pm .15 \text{ m}\cdot\text{s}^{-1}$  for low heels and  $1.22 \pm .12 \text{ m}\cdot\text{s}^{-1}$  for high heels, respectively. However, as previously mentioned, no significant differences in speed were found between a 0 cm shoe and heel height and 3.8 cm in Kerrigan et al. (2005).

### Kinematics and Kinetics

Wearing high heels alters many walking characteristics, and previous research has shown significant influences of heel height on walking kinematics. Footwear, specifically footwear with high heel height, leads to several changes at the distal segments and joints. For instance, greater ankle plantar flexion (Stefanyshyn et al., 2000; Ebbeling et al., 1994; de Lateur et al, 1991; Snow & Williams, 1994; Murray et al., 1970), foot adduction and foot

supination (Stefanyshyn et al; 2000; Snow & Williams, 1994), and rearfoot angle (Snow & Williams, 1994; Ebbeling et al., 1994) have been shown. Greater ankle inversion and foot adduction leads to less pronation, which could affect the shock-attenuation function of pronation (Snow & Williams, 1994). In addition, greater values of maximum ankle plantar flexion, and ankle plantar flexion at foot strike and toe-off as well as decreased ankle range of motion (Snow & Williams, 1994; Stefanyshyn et al, 2000; Murray et al., 1970) have been reported. This corresponds to a reduction in maximum ankle dorsiflexion (Snow & Williams, 1994). Kinetically, increased heel height leads to lower plantar flexion moments (Esenyel et al., 2003; Stefanyshyn et al., 2000; Kerrigan et al., 1998), less plantar flexion work (Esenyel et al., 2003), and lower external ankle eversion moments (Kerrigan et al., 1998).

Previous research provides evidence that ground reaction forces (GRFs) are altered during high-heeled walking. Greater vertical GRFs (Stefanyshyn et al., 2000; Ebbeling et al., 1994; Snow & Williams, 1994; Hong et al., 2005), anterior-posterior GRFs (Stefanyshyn et al., 2000; Snow & Williams, 1994; Hong et al., 2005), and vertical impact forces (Stefanyshyn et al., 2000; Yung-Hui & Wei-Hsien, 2005; Hong et al., 2005) have been shown. Hong et al. (2005) found impact force to significantly increase from  $0.44 \pm .04$  BW in 0 cm shoes to  $.57 \pm .07$  in 7.6 cm heels. As previously mentioned, this change is important because higher impact forces suggests greater shock to be transmitted to more proximal joints, such as the knee and hip. In addition, greater vertical impact loading rate has been reported (Stefanyshyn et al., 2000), which may also be potentially harmful. Snow & Williams (1994) found a prolongation of maximum vertical force and that maximum and minimum anterior-posterior forces occurred earlier in the support phase with higher heel

heights. Pressure changes at the foot have also been shown. For example, greater medial forefoot pressure (Yung-Hui & Wei-Hsien, 2005; Hong et al., 2005; Eisenhardt et al., 1996; Mandato & Nester, 1999), less pressure in the heel and midfoot (Yung-Hui & Wei-Hsien, 2005; Hong et al., 2005), and a shift of pressure to the first metatarsal and hallux (Eisenhardt et al., 1996; Mandato & Nester, 1999) have been reported.

Evident changes occur at the distal segments and joints of the lower-extremity with higher heel heights, which are accompanied by changes at more proximal joints. The most predominant and emphasized kinematic change at the knee is greater knee flexion during stance (Opila-Correia, 1990a; 1990b; Stefanyshyn et al., 2000; de Lateur et al., 1991; Ebbeling et al., 1994; Kerrigan et al., 1998; Kerrigan et al., 2005; Murray et al., 1970). In fact, Ebbeling et al. (1994) showed that maximum knee flexion increased from  $21.1 \pm 6.0^\circ$  at 1.25 cm heel height to  $24.9 \pm 7.1^\circ$  at 7.62 cm heel height. According to Kerrigan et al. (2005), the greater knee flexion during walking contributes to a greater and prolonged external knee flexor moment, which the authors suggest is a key factor in the degeneration of the patellofemoral compartment. The following kinetic changes have been reported: higher knee extensor moment (Esenyel et al., 2003; Stefanyshyn et al., 2000), higher external knee adduction moment (Esenyel et al., 2003; Kerrigan et al., 1998; Kerrigan et al., 2005), and prolonged external knee flexor moment (Kerrigan et al., 1998; Kerrigan et al., 2005).

Alterations in the kinematics and kinetics at the distal segments that occur due to increased heel height lead to further changes at the hip and trunk. Esenyel et al. (2003) and Kerrigan et al. (1998) have demonstrated higher hip extension moments with increased heel height. In addition, Opilia-Correia (1990b) reported greater trunk extension and anterior pelvic tilt angles in young women during high-heeled walking. Although Lee et al. (2001)

did not report greater trunk extension, they reported a decrease in trunk flexion with an increase in heel height. In contrast, Franklin, Chenier, Brauninger, Cook, & Harris (1995) reported increased trunk flexion with higher heel heights statically. Lastly, Lee et al. (2001) and Snow and Williams (1994) reported displacements of the center of mass in the vertical and anterior directions, respectively.

There are clear influences of heel height on the kinematics and kinetics of high-heeled walking. These alterations could be potentially adverse, which has already been suggested in the case of higher impact force and shock transmission through the lower extremity and higher external knee adduction moment, which are thought to contribute to the joint degeneration development and progression of knee OA.

#### Muscle Contribution

Due to alterations in the kinematics and kinetics of high-heeled walking, there are certain changes that also occur in muscle contribution of the lower extremity and trunk. Because higher external knee adduction moments at the knee are evident during high-heeled gait, there must be a counteractive internal abduction moment at the knee generated by soft tissue. Schipplein and Andriacchi (1991) showed that co-contraction of the quadriceps and hamstrings in addition to the possible tension in lateral soft tissues were required to balance the external knee adduction moments present during normal walking. Schipplein and Andriacchi (1991) also found that increased co-activation of the quadriceps and hamstrings could support higher external knee adduction moments, suggesting that the muscles at the knee may protect the medial compartment from greater loads (Bennell, Hunt, Wrigley, Lim, & Hinman, 2008). Similarly, the contribution of the lower extremity muscles and possible effects on repetitive impulsive loading have also been investigated. Because soft tissue is



viscous, it can contribute to shock absorption throughout the lower extremity (Wosk & Voloshin, 1981). In contrast to the skeletal system, soft tissue is less rigid and will undergo lower accelerations at heel strike, which should result in reduced impact forces (Nigg, Cole, & Bruggemann, 1995). Therefore, Nigg et al. (1995) suggests that increased muscle activation in the lower extremity results in a more rigid musculoskeletal system and higher impact forces at heel strike. This was evident in Potthast, Bruggemann, Lundberg, and Arndt (2010) reporting higher impact forces with greater percentage of maximal voluntary contractions of the gastrocnemius, quadriceps, and hamstring muscles when being externally impacted at the heel. However, contrary to Nigg et al. (1995), greater muscle activation also resulted in lower tibial accelerations, meaning increased stiffness of the lower extremity resulted in lower accelerations. On the other hand, Flynn, Holmes, and Andrews (2004) reported a reduction in peak tibial accelerations when subjects were externally impacted at the heel after fatiguing the tibialis anterior and gastrocnemius. They suggested the lower peak tibial accelerations occurred due to a decreased force producing capacity of the muscles, which resulted in the muscles being less stiff than a pre-fatigue state.

In addition, muscle contribution prior to heel strike may also influence impulsive forces at heel strike. Jefferson, Collins, Whittle, Radin and O'Conner (1990) reported that paralysis of the quadriceps muscle resulted in a significant increase in impact force, suggesting that quadriceps activity may be important in decelerating the limb prior to heel contact. Similarly, Verdini, Marcucci, Benedetti, and Leo (2006) found that subjects that did not exhibit a heel strike transient during walking activated the quadriceps and hamstrings muscles prior to heel strike. Subjects that exhibited a heel strike transient during walking typically had an altered activation pattern in one or more of the lower extremity muscles with

the rectus femoris, vastus medialis, and tibialis anterior being the most common. Verdini et al. (2006) concluded that quadriceps and hamstrings muscles were important in limb deceleration and stabilization prior to heel strike and that any alteration in this activation timing may result in a high velocity impact with the ground. Overall, if muscle contribution is altered during walking in higher heel heights, the muscles' role in protecting the knee joint from increased loading may be compromised.

Previous research on muscle contribution in response to increased heel height is limited but does show some evident changes. Stefanyshyn et al. (2000) provided evidence of increased rectus femoris, soleus and peroneus longus activity, and Lee et al. (2001) showed increases in erector spinae muscle activity. However, these studies reported no significant differences in vastus medialis, biceps femoris, semitendinosus, gastrocnemius, and tibialis anterior and vastus lateralis activity. In addition, Edwards, Dixon, Kent, Hodgson, and Whittaker (2008) sought to provide evidence of increased medial loading at the knee by demonstrating greater activity of the vastus medialis (VM) relative to the vastus lateralis (VL) during standing on cork wedges of increasing height. Although increasing heel height from barefoot to 5 cm led to significant increases in the average rectified value for VM from  $84.5 \pm 53.7$  mV to  $102.5 \pm 60.5$  mV and  $52.5 \pm 28.0$  mV to  $62.0 \pm 29.2$  mV, ratios between the activity of the muscles were not significantly different for increased heel height conditions. Overall, previous research on the effect of heel height on muscle contribution is limited, making it difficult to draw conclusions on how it may affect loading on the knee.

### Conclusion

Previous research on high-heeled walking demonstrates changes in the kinematics and kinetics of walking, suggesting the presence of adverse effects. Wearing high heels

affects the body in many ways. For example, a link between wearing high heels and the development and progression of chronic conditions, such as joint degeneration and knee OA has been proposed (Kerrigan et al., 1998; Kerrigan et al., 2005; Voloshin & Loy, 1994). It is evident that frontal plane kinetics and repetitive impulsive loading play an important role in the process of joint degeneration. Specifically, in the frontal plane, Kerrigan et al. (1998) and Kerrigan et al. (2005) have emphasized the possible contributions of high-heeled walking to knee OA through higher peak external knee varus moments (i.e. external knee adduction moments). However, each study only used two conditions and did not investigate the effect of systematically increasing heel height. In addition, although it has been suggested that the repetitive impulsive loading at heel strike during walking is a key mechanical factor in joint degeneration, research investigating high-heeled walking has almost completely ignored this phenomenon. Voloshin and Loy (1994) investigated impact acceleration at several heel heights, but a true systematic manipulation of heel height was not included and there was little control over shoe design. Lastly, both Kerrigan et al. (1998) and Kerrigan et al. (2005) and Voloshin and Loy (1994) only investigated heel heights up to approximately 6 cm and 7 cm, which does not encompass the full range of heel heights typically worn by women. Therefore, further investigation of kinetic variables in the frontal plane and repetitive impulsive loading and the effect of systematically increasing heel heights across a wide range of heights is warranted.

Overall, it is evident that heel height alters characteristics of walking and that these alterations may adversely affect the body and contribute to the development of chronic conditions, such as joint degeneration and knee OA. This is of great importance in that the prevalence of knee OA is increasing and many women wear high heels.

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## CHAPTER 2. THE EFFECT OF HEEL HEIGHT ON FRONTAL PLANE JOINT MOMENTS OF THE LOWER EXTREMITY DURING WALKING

A paper to be submitted to *Gait and Posture*

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### **Abstract**

Wearing high heels alters walking kinematics and kinetics and can create potentially adverse effects on the body. Our purpose was to determine how heel height affects frontal plane joint moments at the hip, knee, and ankle, with a specific focus on the external knee moment due to its importance in medial loading and knee osteoarthritis. Fifteen women completed overground walking trials in three different heel height conditions (1, 5, and 9 cm) for a fixed walking speed ( $1.3 \text{ m}\cdot\text{s}^{-1}$ ) and a preferred walking speed while kinematic and force platform data were collected concurrently. For both fixed and preferred speeds, peak external knee adduction moment increased systematically with an increase in heel height. Peak internal ankle eversion moment increased systematically with an increase in heel height, and though little change was evident at the hip, peak hip abduction moment was slightly higher for the 9 cm heel height compared to the 1 and 5 cm conditions. The higher peak external knee adduction moment with increasing heel height suggests greater medial loading at the knee, which can contribute to joint degeneration and the development of knee osteoarthritis. The kinetic changes observed at the ankle with increasing heel height may contribute to larger medial loads at the knee. Overall, wearing high heels, particularly those with higher heel heights, may put individuals at greater risk for joint degeneration and developing knee OA.

## **Introduction**

Footwear design properties, including cushioning, stiffness, heel width and height, can alter the kinematics and kinetics of walking. A commonly worn type of women's footwear that dramatically alters walking characteristics is high-heeled shoes. Despite a reported lack of comfort and support (Hong, Lee, Chen, Pei, & Wu, 2005; Yung-Hui & Wei-Hsien, 2005), the American Podiatric Medical Association reported over 72% of women wear high heels, with 40% of these women wearing them daily (American Podiatric Medical Association, 2003, as cited in Yoon, An, Yoo, & Kwon, 2009).

Many investigators have reported modifications in walking kinematics and kinetics when wearing high heels. Kinematically, higher heel heights contribute to slower self-selected walking speeds, shorter stride lengths (Opila-Correia, 1990a; Esenyel, Katlen, Walden, & Gitter, 2003; Murray, Kory, & Sepic, 1970) and greater knee flexion, plantar flexion, trunk extension, and anterior pelvic tilt (Ebbeling, Hamill, & Crussemeyer, 1994; Opila-Correia, 1990b; Murray et al., 1970; Snow & Williams, 1994). Kinetically, greater peak vertical and anterior-posterior ground reaction forces throughout stance, medial forefoot pressures, peak knee extensor moments, peak external knee adduction moments, and lower plantar flexion moments have been observed during walking in high heels (Esenyel et al., 2003; Hong et al., 2005; Snow & Williams, 1994; Nyska, McCabe, Linge, & Klenerman, 1999; Kerrigan, Todd, & Riley, 1998; Kerrigan et al., 2005). These observed alterations in kinematics and kinetics are thought to create potentially adverse loading at the joints of the lower extremity, which may be an important factor in the development and progression of chronic conditions (Kerrigan et al., 1998; Kerrigan et al., 2005; Voloshin & Loy, 1994; Nyska et al., 1999; Corrigan, Moore, & Stephens, 1993). Specifically, previous research has

proposed wearing high heels contributes to joint degeneration and the development of knee osteoarthritis (OA) through repetitive dynamic loading during walking (Kerrigan et al., 1998; Kerrigan et al., 2005; Voloshin & Loy, 1994).

Because knee OA occurs most often in the medial tibiofemoral compartment of the knee (Dearborn, Eakin, & Skinner, 1996; Keyes, Carr, Miller, & Goodfellow, 1992), recent research on knee OA has focused on the knee in the frontal plane and possible loading asymmetries within the joint that contribute to knee OA. Schipplein & Andriacchi (1991) demonstrated that unequal forces are transmitted between the medial and lateral compartments during normal walking and reported medial forces may be as much as 2.5 times greater than those on the lateral side. Higher external knee adduction moments are thought to contribute to larger medial compartment loads (Baliunas et al., 2002; Schipplein & Andriacchi, 1991) and are believed to be an important component in the development of knee OA (Sharma et al., 1998; Baliunas et al., 2002; Schipplein & Andriacchi, 1991).

Previous research on high-heeled walking has shown higher external knee adduction moments compared to walking in lower heel heights. Kerrigan et al. (1998) reported 23-24% higher peak external knee varus moment (external knee adduction moment) during walking in 6 cm heels compared to barefoot walking. Peak moments were higher in both early stance (0.32 vs. 0.26  $\text{N}\cdot\text{m}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$  for heels and barefoot conditions, respectively) and late stance (0.26 vs. 0.21  $\text{N}\cdot\text{m}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$ ). In a follow-up study using wide-heeled, moderate height (3.8 cm) shoes, Kerrigan et al. (2005) found peak external knee varus moment during late stance was 14% higher for young women and 9% higher for elderly women for heeled shoes compared to the same style of shoes with 0 cm heels. In addition, Esenyel et al. (2003)

reported peak external knee adduction moment was 25% higher in a wide-heeled shoe of 6 cm heel height compared to a sport shoe with 1 cm heel height.

To our knowledge, the effect of heel height on external knee adduction moment has not been investigated thoroughly and systematically. Kerrigan et al. (1998; 2005) employed only two heel height conditions in their studies. In addition, they allowed subjects to walk at self-selected speeds for each footwear and heel height condition. Although a preferred walking velocity avoids imposing constraints on gait, enhancing real-world application, a fixed walking velocity controls for the speed effect on walking kinematics and kinetics. Lastly, both studies (Kerrigan et al., 1998; 2005) only investigated heel heights up to approximately 6 cm, which does not encompass the full range of heel heights typically worn by women. Additionally, with the exception of Kerrigan et al. (2005), little control over footwear designs compared was exercised in previous studies (Kerrigan et al., 1998; Esenyel et al., 2003), which could potentially confound the effects of heel height.

The purpose of this study was to determine the effect of heel height on net joint moments at the hip, knee, and ankle in the frontal plane, with particular attention placed on the external knee moment. It was hypothesized that frontal plane net joint moments of the lower extremity, specifically peak external knee adduction moment, increase systematically as heel height increases.

## **Methods**

### **Subjects**

Fifteen healthy women participated in this study (age:  $23.8 \pm 4.4$  yrs, height:  $165.5 \pm 7.1$  cm, mass:  $60.9 \pm 8.7$  kg). Subjects were free from injury for at least 12 months prior to participation and were recruited from the community by word of mouth, posted flyers, and

electronic advertisements. Institutional Review Board approval was obtained for this study, and all subjects were provided written informed consent prior to participation.

### Data Collection

Subjects completed two sessions. The first test session was used to orient subjects to the testing protocol and for determination of preferred walking speed for each of the three heel height conditions. The second session served as the primary experimental session in which subjects completed walking trials for each heel height condition at both preferred and fixed ( $1.3 \text{ m}\cdot\text{s}^{-1}$ ) velocities. Both preferred and fixed walking velocity conditions were included in the experiment design because of advantages each offers to understanding heel height effects on gait. Use of preferred walking velocities for different heel height conditions enhanced the ecological validity of the experiment. That is, the effects of heel height on gait characteristics and lower extremity loading at preferred walking velocities, which tend to decrease as heel height increases (Opila-Correia, 1990a; Esenyel et al., 2003; Murray et al., 1970), are more representative of what may be occurring in real life. Walking velocity, however, is known to affect many kinematic and kinetic gait descriptors. Thus, examining the effects of heel height on gait characteristics under controlled or fixed velocity conditions limits the confounding influences of walking velocity and enhances internal validity. For the fixed velocity condition,  $1.3 \text{ m}\cdot\text{s}^{-1}$  was selected based upon previous research including both fixed and preferred walking velocities. Hong et al. (2005) and Yung-Hui and Wei-Hsien (2005) used the fixed walking velocity of  $1.3 \text{ m}\cdot\text{s}^{-1}$ , while fixed velocities from other previous research ranged from  $1.11 \text{ m}\cdot\text{s}^{-1}$  (Lee, Jeong, & Freivalds, 2001; Nyska et al., 1996) to  $1.4 \text{ m}\cdot\text{s}^{-1}$  (Snow & Williams, 1994; Stefanyshyn, Nigg, Fisher, O'Flynn & Liu, 2000). Preferred walking velocities while wearing high heels that have been reported in the

literature have ranged from  $1.23 \text{ m}\cdot\text{s}^{-1}$  (Opilia-Correia, 1990b) to  $1.44 \text{ m}\cdot\text{s}^{-1}$  (de Lateur, Giaconi, Questad, Ko, & Lehmann, 1991).

Session 1. Prior to participation, subjects were familiarized with the experimental protocol. In addition, subjects completed a short questionnaire assessing high heel experience. The questionnaire was used to obtain descriptive information on frequency, duration, and consistency of high heel usage. Subjects were fitted in black spandex clothing to facilitate reflective marker positioning on the body and their identification during data analysis. Body weight, body height, and anthropometric measurements of the right leg and foot were obtained. The following anthropometric measurements were taken: foot, shank, and thigh length, shank and thigh maximum circumference, foot width, malleolus width, and malleolus height.

Subjects were fitted with three pairs of shoes with heel heights of 0.8, 5.1, and 8.9 cm (hereafter referred to as 1, 5, and 9 cm heel heights). All shoes, which carried the Nine West brand name, were commercially available and selected based upon their similarity of construction so that the main difference amongst shoes was heel height (Figure 1). Shoe construction included a narrow heel, a narrow forefoot and toe box, and limited cushioning material. Shoe sizes included 6, 7, 8, and 9.



Figure 1. Nine West shoes of heel height 1, 5, and 9 cm.

Preferred walking velocity was then measured for each of the three heel height conditions. The order of the heel heights was randomly assigned for each subject. One reflective marker was placed on the low back over the joint between the fifth lumbar and first sacral (L5S1) vertebrae. After at least a 5-minute walking period in each heel height condition, subjects walked overground through the measurement zone of the Vicon motion system (Vicon, Centennial, Colorado) at a comfortable pace while contacting a force platform (AMTI, Watertown, MA) in the center of the walkway with the right foot. Subjects were explicitly encouraged to use a comfortable speed as if they were walking from their car to the front door of the grocery store. Following several practice trials, the L5S1 reflective marker position was sampled at 200 Hz and force platform was sampled at 1000 Hz for 10 preferred speed trials. The average forward horizontal velocity of the reflective marker over



the period of contact with the force platform was computed for each trial. The average of the 10 trials represented the subject's preferred walking velocity for a given heel height.

To conclude the first test session, subjects practiced walking at  $1.3 \text{ m}\cdot\text{s}^{-1}$  for each of the three heel height conditions. Subjects completed as many practice trials as necessary until their walking speed fell consistently within 3% of the target speed.

Session 2. The second session was completed within 7 days of session one. Subjects again wore black spandex clothing. Reflective markers were placed on the skin, clothing, or shoes over anatomical landmarks of the subject's lower trunk, pelvis, and right lower extremity. Markers were attached over the dorsal aspect of the foot slightly posterior to the third metatarsal head, 5<sup>th</sup> metatarsal head, heel, medial and lateral malleolus, proximal and distal anterior calf, posterior calf, medial and lateral femoral epicondyles, lateral and anterior thigh, left and right greater trochanter, left and right anterior superior iliac spine, and the joint between the fifth lumbar and first sacral (L5S1) vertebrae. Medial malleolus and medial femoral epicondyle markers were removed for dynamic trials (Figure 2).



Figure 2. Reflective marker placement.

Subjects were provided with the same shoes worn in session one and completed heel height conditions in the same order as session one. Prior to each heel height condition, subjects completed as many practice trials as needed to consistently match the target velocity while contacting a force platform in the center of the walkway with the right foot. Ten acceptable trials for each of the three heel height conditions were completed at the preferred walking velocities measured in test session one. Subjects then completed ten acceptable trials for each of the three heel height conditions at  $1.3 \pm 0.04 \text{ m}\cdot\text{s}^{-1}$ . Acceptable trials were those in which average horizontal velocity of the L5S1 marker during the stance phase on the force platform fell within 3% of the target velocity, the right foot was completely on the

force platform, and there was no visible indication the participant altered step characteristics to impact the force platform.

During each walking trial, marker position and ground reaction force (GRF) data were collected synchronously using an eight-camera Vicon motion system. The motion analysis system was calibrated using a two step process: 1) a rigid reference frame of known coordinates was waved throughout the cameras' view to establish a calibrated volume of space, and 2) the reference frame was used to establish the origin and direction of the Cartesian coordinate system. Marker position and GRF data were sampled at 200 Hz and 1000 Hz, respectively.

### Data Analysis

The raw marker position and force platform data were imported into Matlab (The Mathworks, Natick, MA) for data processing. A 4<sup>th</sup> order zero-lag Butterworth filter with a low-pass cutoff frequency of 6 Hz was used to smooth raw marker position (Winter, 1990) and force platform data (Bisseling & Hof, 2006). The right leg stance phase was analyzed for all kinematic and kinetic variables with the exception of stride length and stride rate. To define the stance phase, a vertical ground reaction force threshold of 20 N was used for the identification of heel strike and toe off events. Stride length was calculated as the difference in the position of the heel marker on the right foot along the anterior-posterior axis of the body for consecutive heel strikes of the right foot throughout the data collection period. Stride rate was computed as the inverse of stride time, which was determined from the time between consecutive heel strikes of the right foot. Marker position data were used to calculate three-dimensional segment angles for the pelvis, thigh, shank, and foot. Segment and joint angles were calculated using marker clusters to create rotation matrices for the

pelvis, thigh, shank, and foot using Cardan angles with a flexion/extension, abduction/adduction, internal/external rotation sequence. Segment angular velocities and accelerations and segment center of mass linear velocities and accelerations were calculated using finite difference approximation methods.

Anthropometric measurements were used to estimate segment masses, segment center of mass locations, and segment moments of inertia needed for joint moment calculations (Vaughan et al., 1992). Every 5<sup>th</sup> value in the ground reaction force data set was extracted to correspond to marker position sampling frequency. Net internal three-dimensional joint moments were calculated using inverse dynamics with rigid body assumptions (Vaughn et al., 1992). Because of the limited musculature counteracting the external knee adduction moment (Schipplein & Andriacchi, 1991), the frontal plane knee moment was presented as an external knee moment, whereas the hip and ankle was expressed as internal joint moments. Net joint moments were normalized in amplitude to body mass (Nm/kg). Joint moments were also normalized in time to percentage of stance phase in 1% increments so that ensemble averages of moment profiles could be computed.

### Statistical Analysis

A one-way analysis of variance (ANOVA) with repeated measures was used to assess the effect of heel height on lower extremity frontal plane net joint moments and walking kinematics for both preferred and fixed walking velocity conditions. The means of ten trials for each subject for each heel height condition were used in the statistical analysis. The alpha significance level was equal to 0.05. In the event of a significant heel height effect, a post hoc Bonferroni adjustment ( $p = 0.05/3 = 0.017$ ) was used to assess pairwise comparisons.

## Results

Subjects reported wearing high heels an average of 2.1 (SD = 1.6) days per week, ranging from once every six months to 5 days per week. The time spent wearing high heels averaged 5.1 (SD = 1.6) hours per episode. The typical heel height worn by subjects averaged 7.6 cm (SD = 1.5, range: 5.1 – 10.2 cm). Mean length of time of consistently wearing high heels was 7.6 years (SD = 4.4, range: 7 months to 20 years).

Walking velocity was effectively controlled for the fixed speed trials. Subjects were able to closely match the  $1.3 \text{ m}\cdot\text{s}^{-1}$  target speed (Table 1). For the preferred speed condition, subjects chose a significantly slower preferred walking velocity for the 9 cm heel height compared to the 1 and 5 cm conditions ( $F_{2,13} = 7.34$ ,  $p = .003$ ) (Table 1). Preferred speeds were similar to those previously reported in heel height investigations, which ranged from  $1.23 \text{ m}\cdot\text{s}^{-1}$  (Opilia-Correia, 1990b) to  $1.44 \text{ m}\cdot\text{s}^{-1}$  (de Lateur et al., 1991).

For fixed speed trials, stride length (SL) systematically decreased ( $F_{2,13} = 13.71$ ,  $p < .001$ ) and stride rate (SR) increased ( $F_{2,13} = 7.18$ ,  $p = .003$ ) as heel height increased. The SL and SR values are consistent with those reported in previous research (Opilia-Correia, 1990a; Esenyel et al., 2003; Murray et al., 1970). For preferred speed trials, SL was significantly shorter for the 9 cm condition compared to the 1 and 5 cm heel heights ( $F_{2,13} = 13.61$ ,  $p < .001$ ). SR was not affected by heel height changes for preferred walking velocity trials ( $F_{2,13} = 0.09$ ,  $p = .91$ ). The shorter SL for the 9 cm heel height condition accounted for the slower walking velocity for that condition.

Table 1. Mean (SD) kinematic variables in response to heel height at fixed and preferred speeds.

	Fixed Speed			Preferred Speed		
	1 cm	5 cm	9 cm	1 cm	5 cm	9 cm
<b>V (m·s<sup>-1</sup>)<sup>b</sup></b>	1.31(0.01)	1.30(0.01)	1.30(0.01)	1.35(0.10)	1.35(.15)	1.28(.15)
<b>SL (m)<sup>a b</sup></b>	1.40(0.07)	1.37(0.07)	1.34(0.06)	1.40(0.09)	1.40(0.09)	1.33(0.11)
<b>SR (stride·s<sup>-1</sup>)<sup>a</sup></b>	0.94(0.05)	0.95(0.06)	0.97(0.04)	0.97(0.02)	0.96(0.02)	0.97(0.02)

V = walking velocity, <sup>a</sup> = significant heel height effect for fixed speed trials, <sup>b</sup> = significant heel height effect for preferred speed trials.

The net hip moments in the frontal plane during stance were abduction moments for nearly all of the stance phase (Figure 3, panels A and B), which is consistent with preventing pelvic drop during walking. The profile and amplitude of the net frontal plane hip moments were consistent with those reported by Esenyel et al. (2003). Although similar heel height effects were observed for both fixed and preferred speed conditions, only the peak hip abduction moment during early stance for the fixed speed condition reached statistical significance ( $F_{2,13} = 17.02$ ,  $p < .001$ ). The peak hip abduction moment for the 9 cm heel height condition was approximately 10% higher than peak abduction moments for the 1 and 5 cm heel heights. Heel height did not significantly affect the second peak hip abduction moment that occurs in late stance (fixed:  $F_{2,13} = 2.34$ ,  $p = .115$ ; preferred speed:  $F_{2,13} = .19$ ,  $p = 0.83$ ).

As expected, the net external knee moment reflected varus loading on the joint during most of the stance phase (Figure 3, panels C and D), as has been reported in previous heel height studies (Kerrigan et al., 1998; Kerrigan et al., 2005; Esenyel et al., 2003). Consistent with our hypothesis, peak external knee adduction moments early and late in the stance phase increased systematically as heel height increased for both fixed speed (early stance:  $F_{2,13} =$

8.71,  $p = .001$ ; late stance:  $F_{2,13} = 20.77$ ,  $p < .001$ ) and preferred speed conditions (Figure 4; early stance:  $F_{2,13} = 3.96$ ,  $p = .03$ ; late stance:  $F_{2,13} = 26.79$ ,  $p < .001$ ).

Net frontal plane ankle moments, which were similar in amplitude and pattern to those reported previously (Monaghan, Delahunt, & Caulfield, 2006; Kerrigan et al., 1998), were dramatically affected by changes in heel height. Under the 1 cm heel height condition, the net internal ankle moment was a low amplitude inversion moment during the first 80% of stance before becoming eversion in nature during late stance. Under the 5 and 9 cm heel height conditions, the net ankle moment was an eversion moment for nearly the entire stance phase. The peak ankle eversion moment in late stance for both fixed and preferred speed conditions increased systematically as heel height increased (Figure 3, panels E and F; fixed:  $F_{2,13} = 16.00$ ,  $p < .001$ ; preferred:  $F_{2,13} = 15.79$ ,  $p < .001$ ). Specifically, peak ankle eversion moment for the 5 cm heel height was 64 to 70% higher than the 1 cm condition, while peak eversion moment for the 9 cm heel height was approximately 107% higher than the 1 cm condition.

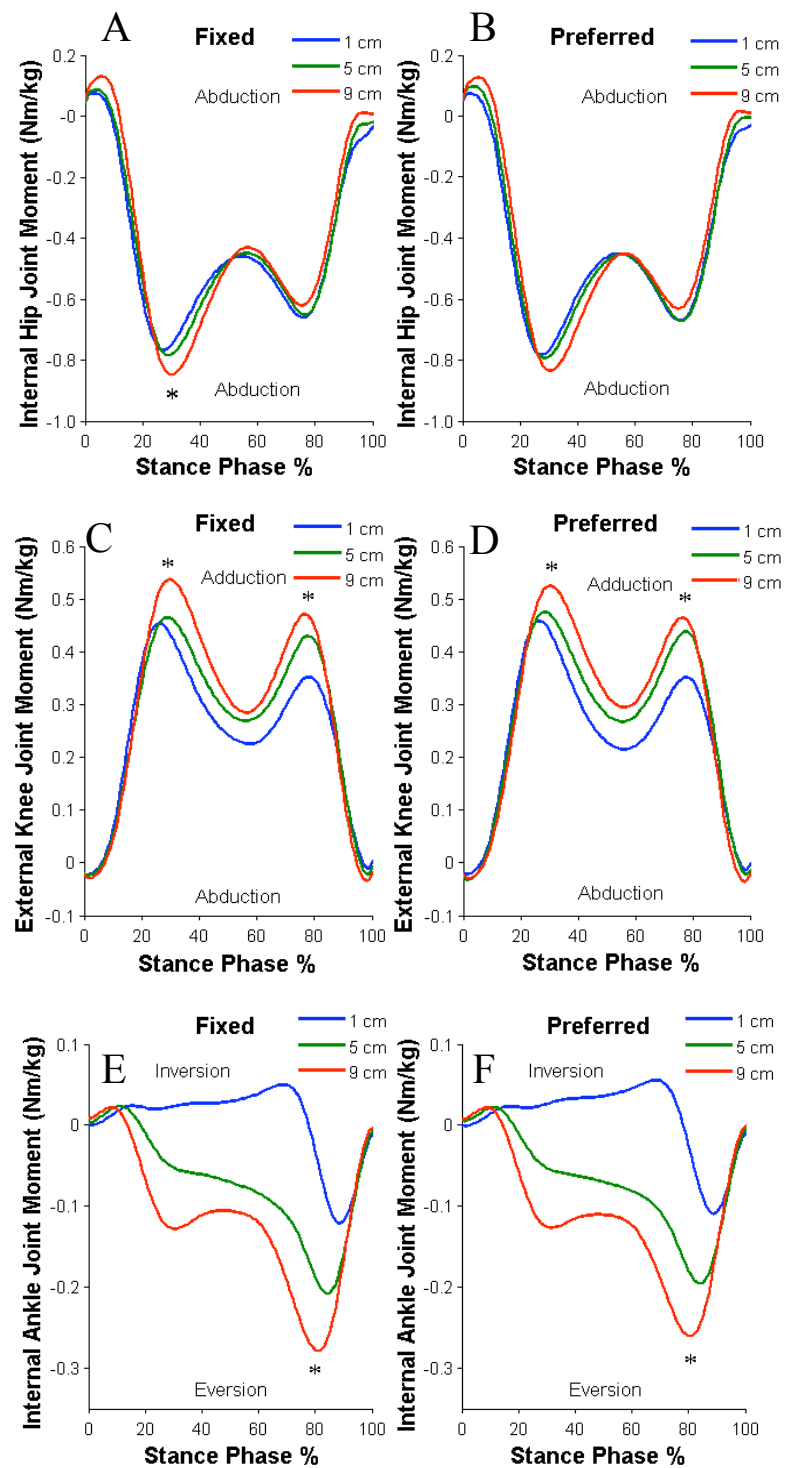


Figure 3. Ensemble averages for frontal plane hip (A, B), knee (C, D), and ankle (E, F) moments in response to heel height for fixed and preferred speed conditions. (\* significant heel height effect).



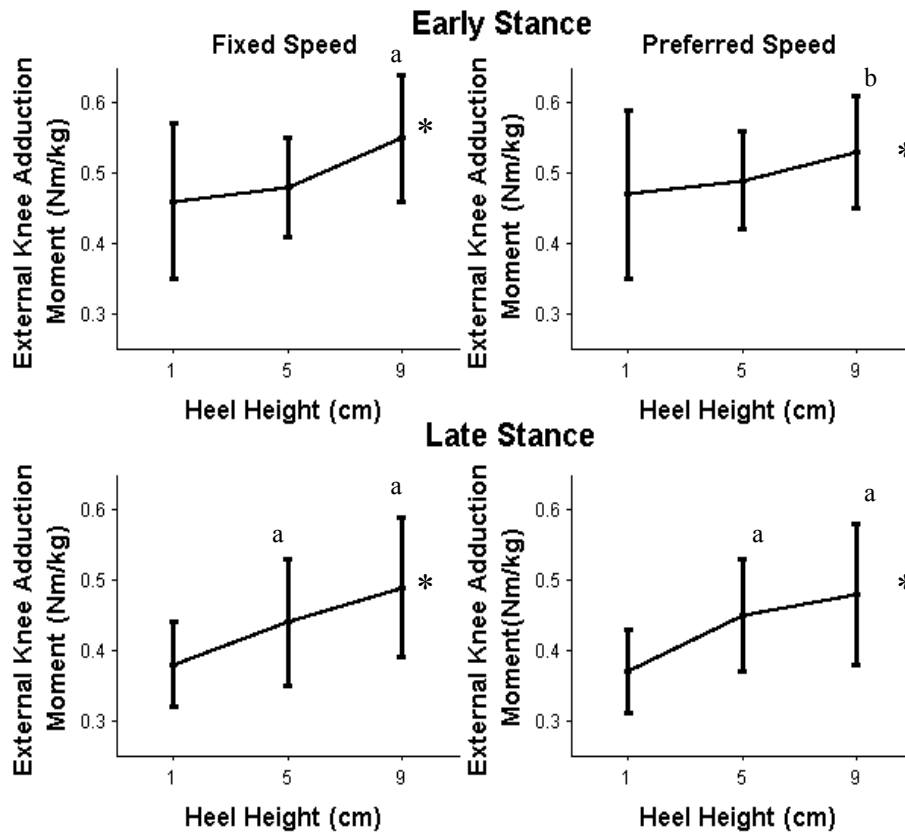


Figure 4. The peak knee adduction moments both early and late in the stance phase of walking increased as heel height increased. These trends were reflected under both fixed speed and preferred speed conditions (\* significant heel height effect, a = significantly different from 1 cm, b = significantly different from 5 cm).

## Discussion

### External Knee Moment

As hypothesized, peak external knee adduction moments systematically increased as heel height increased. These results are consistent with previous heel height investigations (Kerrigan et al., 1998; Kerrigan et al., 2005; Esenyel et al., 2003). Kerrigan et al. (1998) found a 23% higher peak external knee varus moment (i.e., external knee adduction moment) in shoes with 6 cm heels compared to barefoot walking in both early and late stance.

Similarly, Kerrigan et al. (2005) observed higher peak external knee varus moment for a 3.8 cm heel height, wide-heeled shoe compared to a 0 cm shoe. Esenyel et al. (2003) reported peak external knee adduction moment was 25% higher in a 6 cm high, wide-heeled shoe compared to a sport shoe with 1 cm heel height. Our results demonstrate that the direct relationship between heel height and peak external knee adduction moment can be extended beyond moderate heel heights, which has important implications on how higher heel heights affect the medial loading in the knee.

Higher external knee adduction moments contribute to larger loads on the medial compartment of the knee (Baliunas et al., 2002; Schipplein & Andriacchi, 1991). Knee osteoarthritis (OA) occurs most often in the medial compartment (Dearborn et al., 1996; Mow et al., 1995; Keyes et al., 1992). Thus, high external knee adduction moments are believed to be an important contributor to the development of knee OA (Sharma et al., 1998; Baliunas et al., 2002; Schipplein & Andriacchi, 1991). Our results demonstrated peak external knee adduction moments, and consequently, medial loading on the knee, increased systematically as heel height increased. Therefore, wearing high heels consistently, especially higher heels, may put individuals at greater risk for joint degeneration and the development of knee OA.

#### Internal Ankle Moment

Also in support of our hypothesis, ankle frontal plane moments increased in amplitude as heel height increased. Specifically, the peak ankle eversion moment was significantly higher for the 5 and 9 cm heel heights compared to the 1 cm heel height for both fixed and preferred speed conditions. In addition, the inversion moment at the ankle observed for the 1 cm condition was not observed in the 5 and 9 cm conditions. These

results are consistent with Kerrigan et al. (1998) who reported a significant reduction in peak external ankle eversion moment (internal inversion moment) in 6 cm heels compared to barefoot walking.

A factor that likely contributed to the substantial changes in the frontal plane ankle moment profile was the change in the angular orientation of the foot in the frontal plane during the stance phase (Figure 5). Our results indicated the foot segment angle relative to a vertical reference increased as heel height increased for both speed conditions. A positive angle indicated the medial aspect of the foot was positioned superior to the lateral aspect of the foot. As this angle became more positive (i.e., the angular deviation from vertical increased), the external forces acting on the foot attempts to cause inversion of the ankle. This external loading of the foot must be counterbalanced by an internal eversion moment about the ankle.

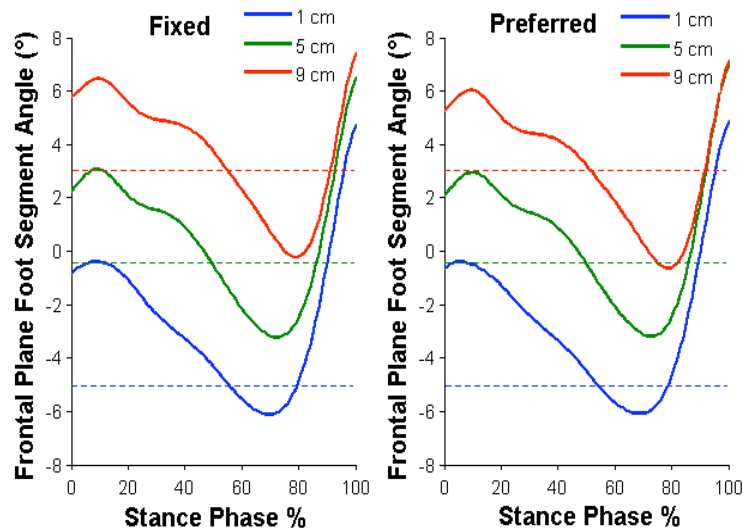


Figure 5. The angle of the foot segment relative to a vertical axis in the frontal plane increased as heel height increased for both fixed and preferred speed conditions. (--) indicate the foot segment angles in a static standing position for each heel height.

### Affect on Knee Loading

The changes in walking kinetics at the ankle due to increasing heel height may also influence loading at the knee. Lateral foot wedges, which typically contribute to a more everted ankle position, have been used to reduce the external knee adduction moment in an effort to reduce the medial loading that contributes to knee OA. Erhart, Mundermann, Mundermann, and Andriacchi (2008) examined the effect of laterally wedging of the foot on frontal plane moments at the ankle and knee. They observed that increases in lateral wedging of the foot by 4 and 8 degrees relative to a neutral shoe increased the peak external ankle eversion moment (i.e., internal inversion moment) by 70 and 131%, respectively, and reduced the peak external knee adduction moment by 8 and 13%, respectively. In similar respect, lower peak external knee adduction moments in late stance have been associated with lower peak external ankle inversion moments (i.e. internal eversion moments) (Andrews, Noyes, Hewett, & Andriacchi, 1996). Taken together, these results suggest that higher peak internal ankle eversion moments contribute to higher external knee adduction moments. Our results, which show both higher peak ankle eversion moments and higher peak external knee adduction moments with increasing heel height, are consistent with these observations. Therefore, kinematic and kinetic alterations at the ankle in walking with higher heel heights may increase medial compressive loading at the knee, which may contribute to medial compartment knee OA.

### Internal Hip Moment

There was limited support for our hypothesis that the frontal plane hip moment increases with increasing heel height. Only the peak abduction moment for the 9 cm heel height in early stance under the fixed speed condition differed significantly from other

values. Esenyel et al. (2003) reported the peak hip abduction moment was 11% higher for a shoe with 6 cm heel compared to one with a 1 cm heel. Kerrigan et al. (1998), however, found no significant changes in the frontal plane hip moment between barefoot and 6 cm heel conditions. With respect to loading at the knee, a higher peak hip abduction moment during stance would attempt to produce a more varus position at the knee, thereby contributing to loading on the medial aspect of the knee. It is important to note again, however, the differences in frontal plane hip moments between heel height conditions were small. Thus, the adaptations at the hip with increasing heel height likely do not contribute substantially to adverse loading at the knee.

Several limitations of this study are worth noting. First, subjects were provided with shoes instead of wearing their own. Although this enhances the internal validity of the study by controlling footwear structure, it also could have potentially created changes in gait characteristics due to subjects not being accustomed to wearing the shoes. Subjects were given adequate time, however, to become familiarized to each shoe condition prior to data collection in order to minimize this response. In addition, the sample reflected considerable variability in the frequency of wearing high heels. The frequency of wearing high heels may affect walking kinematics and kinetics, particularly the variability of gait characteristics. However, a cursory comparison of the gait kinematics and kinetics of the most frequent high heel wearers to those who wear high heels least frequently revealed no obvious differences in response to heel height. In support of this observation, Ebbeling et al. (1994) reported no significant differences in the kinematic and kinetic gait parameters between experienced (3 times/ week; 8 hours/day) and inexperienced (less than 3 times/week; 8 hours/day) wearers of high heels.

Future studies on heel height could investigate the effects of time of exposure to high heels, investigating either walking or standing, in order to understand how the body adapts to higher heel heights over a period of time. Additionally, it may be beneficial to determine if adverse effects, such as the external knee adduction moment, become enhanced over a time of exposure. Further investigation could also include more heel height conditions in order to better define the relationship between heel height and external knee adduction moment as well as other kinematic and kinetic variables.

In conclusion, we found support for our hypothesis that frontal plane net joint moments of the lower extremity during the stance phase of walking, particularly the peak external knee adduction moment, increase systematically as heel height increases. The observed kinetic changes contribute to higher compressive loading on the medial aspect of the knee. This is important as knee OA occurs most often in the medial compartment, and greater loading on this compartment can contribute to degeneration of the joint and the development of knee OA. Therefore, wearing high heels may put individuals at greater risk for developing medial compartment knee OA.

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### CHAPTER 3. THE EFFECT OF HEEL HEIGHT ON IMPACT ACCELERATION AND SHOCK ATTENUATION DURING WALKING

A paper to be submitted to *Gait and Posture*

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#### **Abstract**

During walking, the impact at heel strike creates repetitive impulsive loading of the musculoskeletal system that may contribute to joint degeneration. Our purpose was to determine how heel height affects impact acceleration and shock attenuation during walking. Fifteen women completed overground walking trials in three different heel height conditions (1, 5, and 9 cm) for a fixed walking speed ( $1.3 \text{ m}\cdot\text{s}^{-1}$ ) and a preferred walking speed. Accelerometers were attached to the tibia and forehead in order to measure peak tibial and head acceleration and shock attenuation of the body. Kinematic and force platform data were collected concurrently. It was hypothesized that peak tibial acceleration (PTA) and shock attenuation (SA) increase as heel height increases, while peak head acceleration (PHA) is unaffected by heel height. For both fixed and preferred speeds, PTA was significantly higher for the 5 cm heel height condition compared to the 1 and 9 cm conditions, which were similar in PTA amplitude. PHA followed a similar but more subtle trend as PTA for both speed conditions. Substantial shock attenuation (70 to 75%), which was slightly higher for the 5 cm heel height condition, was responsible for the muted PHA responses in both speed conditions. Although not specifically identified, the unexpected PTA response is likely due to a combination of kinematic gait alterations that result in a lower PTA for the 9 cm heel height. This may have important implications on how heel height affects repetitive impulsive loading during walking.

## Introduction

Footwear design properties, including cushioning, stiffness, heel width and height, can alter the kinematics and kinetics of walking. A commonly worn type of women's footwear that dramatically alters walking characteristics is high-heeled shoes. Despite a reported lack of comfort and support (Hong, Lee, Chen, Pei, & Wu, 2005; Yung-Hui & Wei-Hsien, 2005), the American Podiatric Medical Association reported over 72% of women wear high heels, with 40% of these women wearing them daily (American Podiatric Medical Association, 2003, as cited in Yoon, An, Yoo, & Kwon, 2009).

Many investigators have reported modifications in walking kinematics and kinetics when wearing high heels. Kinematically, higher heel heights contribute to slower self-selected walking speeds, shorter stride lengths (Opila-Correia, 1990a; Esenyel, Katlen, Walden, & Gitter, 2003; Murray, Kory, & Sepic, 1970) and greater knee flexion, plantar flexion, trunk extension, and anterior pelvic tilt (Ebbeling, Hamill, & Crussemeyer, 1994; Opila-Correia, 1990b; Murray et al., 1970; Snow & Williams, 1994). Kinetically, greater peak vertical and anterior-posterior ground reaction forces throughout stance, medial forefoot pressures, peak knee extensor moments, peak external knee adduction moments, and lower plantar flexion moments have been observed during walking in high heels (Esenyel et al., 2003; Hong et al., 2005; Snow & Williams, 1994; Nyska, McCabe, Linge, & Klenerman, 1999; Kerrigan, Todd, & Riley, 1998; Kerrigan et al., 2005). These observed alterations in kinematics and kinetics are thought to create potentially adverse loading at the joints of the lower extremity, which may be an important factor in the development and progression of chronic conditions (Kerrigan et al., 1998; Kerrigan et al., 2005; Voloshin & Loy, 1994; Nyska et al., 1999; Corrigan, Moore, & Stephens, 1993). Specifically, previous research has

proposed wearing high heels contributes to joint degeneration and the development of knee osteoarthritis (OA) through repetitive dynamic loading during walking (Kerrigan et al., 1998; Kerrigan et al., 2005; Voloshin & Loy, 1994).

During normal walking, the body is subjected to impulsive loading at heel strike that is transmitted vertically through the musculoskeletal system in the form of a shockwave (Radin, Paul, & Rose, 1972; Wakeling, Liphardt, & Nigg, 2003). This repetitive impulsive loading is also thought to contribute to joint degeneration (Radin et al., 1972; Wosk & Voloshin, 1981; Collins & Whittle, 1989). The body attenuates the transient shock as it's transmitted up the body in order to minimize accelerations at the head, protecting the visual and vestibular functions needed for postural stability during walking (Voloshin & Wosk, 1982; Ratcliffe & Holt, 1997; Menz, Lord, & Fitzpatrick, 2003). It has been suggested that the body's natural shock absorbers, which include soft tissue, bone, and joint structures such as menisci, articular cartilage, and intervertebral discs (Wosk & Voloshin, 1981; Collins & Whittle, 1989), progressively weaken due to repeated, cyclical loading, allowing for initial damages in the knee joint and further joint degeneration due to compromised protection (Radin et al., 1972; Folman, Wosk, Voloshin, & Liberty, 1986; Voloshin, 1988). Most footwear designs supplement the body's natural shock absorbers by contributing to shock attenuation during walking. The use of compliant insoles in shoe cushioning acts to absorb shock at impact with the ground (Light, McLellan, & Klenerman, 1980; Voloshin, 1988). However, the design of high-heeled shoes limits their ability to attenuate shock during walking. The rigid structure and lack of cushioning capability of most high-heeled shoes allow for less shock absorption at heel strike impact compared to the cushioned sole and insole of most footwear designs (Voloshin & Loy, 1994).

Research investigating the effects of heel height on transient impulsive force at heel strike and shock attenuation throughout the body is limited. To our knowledge, only one study has investigated the effect of heel height on the magnitude of impact accelerations produced at heel strike during walking. In a limited assessment of three subjects, Voloshin and Loy (1994) reported peak tibial accelerations at heel strike ranged from 1.87 to 2.86 g in flat shoes of approximately 1.4 cm heel height to 3.45 to 4.51 g in heels approximately 7 cm in height. Although several heel heights in addition to a barefoot and flat (0.5cm) condition were investigated by Voloshin and Loy, heel heights varied among subjects. Thus, a systematic manipulation of heel height was not completed. In addition, shock attenuation, or the attenuation of the shockwave as it was transmitted up the body, was not measured by Voloshin and Loy. Because the body acts to minimize head acceleration (Menz et al., 2003; Pozzo, Berthoz, & Lefort, 1990), higher tibial accelerations imply a greater need for shock attenuation by the body's natural shock absorbers, which may contribute to their progressive weakening. Lastly, Voloshin and Loy (1994), similar to much of high-heeled walking research, only investigated heel heights up to approximately 7 to 7.5 cm, which does not encompass the full range of heel heights typically worn by women.

Given the limited research on impact acceleration and shock attenuation during walking in high heels, the purpose of this study was to determine the systematic effect of heel height on peak tibial and head axial accelerations and shock attenuation of the body during walking. It was hypothesized that peak tibial acceleration increases as heel height increases. Because of the need to maintain a stable head position during walking, we also predicted shock attenuation increases systematically as heel height increases. Finally, as a result of

greater shock attenuation, we hypothesized that peak head acceleration is unaffected by heel height.

## **Methods**

### Subjects

Fifteen healthy women participated in this study (age:  $23.8 \pm 4.4$  yrs, height:  $165.5 \pm 7.1$  cm, mass:  $60.9 \pm 8.7$  kg). Subjects were free from injury for at least 12 months prior to participation and were recruited from the community by word of mouth, posted flyers, and electronic advertisements. Institutional Review Board approval was obtained for this study, and all subjects were provided written informed consent prior to participation.

### Data Collection

Subjects completed two sessions. The first test session was used to orient subjects to the testing protocol and for determination of preferred walking speed for each of the three heel height conditions. The second session served as the primary experimental session in which subjects completed walking trials for each heel height condition at both preferred and fixed ( $1.3 \text{ m}\cdot\text{s}^{-1}$ ) velocities. Both preferred and fixed walking velocity conditions were included in the experiment design because of advantages each offers to understanding heel height effects on gait. Use of preferred walking velocities for different heel height conditions enhanced the ecological validity of the experiment. That is, the effects of heel height on gait characteristics and lower extremity loading at preferred walking velocities, which tend to decrease as heel height increases (Opila-Correia, 1990a; Esenyel et al., 2003; Murray et al., 1970), are more representative of what may be occurring in real life. Walking velocity, however, is known to affect many kinematic and kinetic gait descriptors. Thus, examining

the effects of heel height on gait characteristics under controlled or fixed velocity conditions limits the confounding influences of walking velocity and enhances internal validity. For the fixed velocity condition,  $1.3 \text{ m}\cdot\text{s}^{-1}$  was selected based upon previous research including both fixed and preferred walking velocities. Hong et al. (2005) and Yung-Hui and Wei-Hsien (2005) used the fixed walking velocity of  $1.3 \text{ m}\cdot\text{s}^{-1}$ , while fixed velocities from other previous research ranged from  $1.11 \text{ m}\cdot\text{s}^{-1}$  (Lee, Jeong, & Freivalds, 2001; Nyska et al., 1996) to  $1.4 \text{ m}\cdot\text{s}^{-1}$  (Snow & Williams, 1994; Stefanyshyn, Nigg, Fisher, O'Flynn & Liu, 2000). Preferred walking velocities while wearing high heels that have been reported in the literature have ranged from  $1.23 \text{ m}\cdot\text{s}^{-1}$  (Opilia-Correia, 1990b) to  $1.44 \text{ m}\cdot\text{s}^{-1}$  (de Lateur, Giaconi, Questad, Ko, & Lehmann, 1991).

Session 1. Prior to participation, subjects were familiarized with the experimental protocol. In addition, subjects completed a short questionnaire assessing high heel experience. The questionnaire was used to obtain descriptive information on frequency, duration, and consistency of high heel usage. Subjects were fitted in black spandex clothing to facilitate reflective marker positioning on the body and their identification during data analysis. Body weight and height were measured, and subjects were fitted with three pairs of shoes with heel heights of 0.8, 5.1, and 8.9 cm (hereafter referred to as 1, 5, and 9 cm heel heights). All shoes, which carried the Nine West brand name, were commercially available and selected based upon their similarity of construction so that the main difference amongst shoes was heel height (Figure 1). Shoe construction included a narrow heel, a narrow forefoot and toe box, and limited cushioning material. Shoe sizes included 6, 7, 8, and 9.



Figure 1. Nine West shoes of heel height 1, 5, and 9 cm.

Preferred walking velocity was then measured for each of the three heel height conditions. The order of the heel heights was randomly assigned for each subject. One reflective marker was placed on the low back over the joint between the fifth lumbar and first sacral (L5S1) vertebrae. After at least a 5-minute walking period in each heel height condition, subjects walked overground through the measurement zone of the Vicon motion system (Vicon, Centennial, Colorado) at a comfortable pace while contacting a force platform (AMTI, Watertown, MA) in the center of the walkway with the right foot. Subjects were explicitly encouraged to use a comfortable speed as if they were walking from their car to the front door of the grocery store. Following several practice trials, the L5S1 reflective marker position was sampled at 200 Hz and force platform was sampled at 1000 Hz for 10 preferred speed trials. The average forward horizontal velocity of the reflective marker over

the period of contact with the force platform was computed for each trial. The average of the 10 trials represented the subject's preferred walking velocity for a given heel height.

To conclude the first test session, subjects practiced walking at the fixed velocity of  $1.3 \text{ m}\cdot\text{s}^{-1}$  for each of the three heel height conditions. Subjects completed as many practice trials as necessary until their walking speed fell consistently within 3% of the target speed.

Session 2. The second session was completed within 7 days of session one. Subjects again wore black spandex clothing. Reflective markers were placed on the skin, clothing, or shoes over anatomical landmarks of the subject's lower trunk, pelvis, and right lower extremity. Markers were attached over the dorsal aspect of the foot slightly posterior to the third metatarsal head, 5<sup>th</sup> metatarsal head, heel, medial and lateral malleolus, proximal and distal anterior calf, posterior calf, medial and lateral femoral epicondyles, lateral and anterior thigh, left and right greater trochanter, left and right anterior superior iliac spine, and the joint between the fifth lumbar and first sacral (L5S1) vertebrae. Medial malleolus and medial femoral epicondyle markers were removed for dynamic trials (Figure 2).

Low mass, uniaxial accelerometers (Analog Devices, model ADXL278, Norwood, MA) were mounted to the skin of the distal medial-anterior portion of the tibia and the center of the forehead (Figure 2). The accelerometers were aligned with the longitudinal axis of the tibia and head. Elastic velcro bands and elastic athletic tape were wrapped around the shank and head to minimize skin movement artifact (Wosk & Voloshin, 1981; Derrick, Hamill, & Caldwell, 1998). The accelerometers and a customized event marker were connected to a data logger (Biomedical Monitoring, model BM42, Glasgow, Scotland) that was attached to the shorts near the left hip of the subject. The event marker was a button held in the subject's left hand and was used to help distinguish trial conditions.



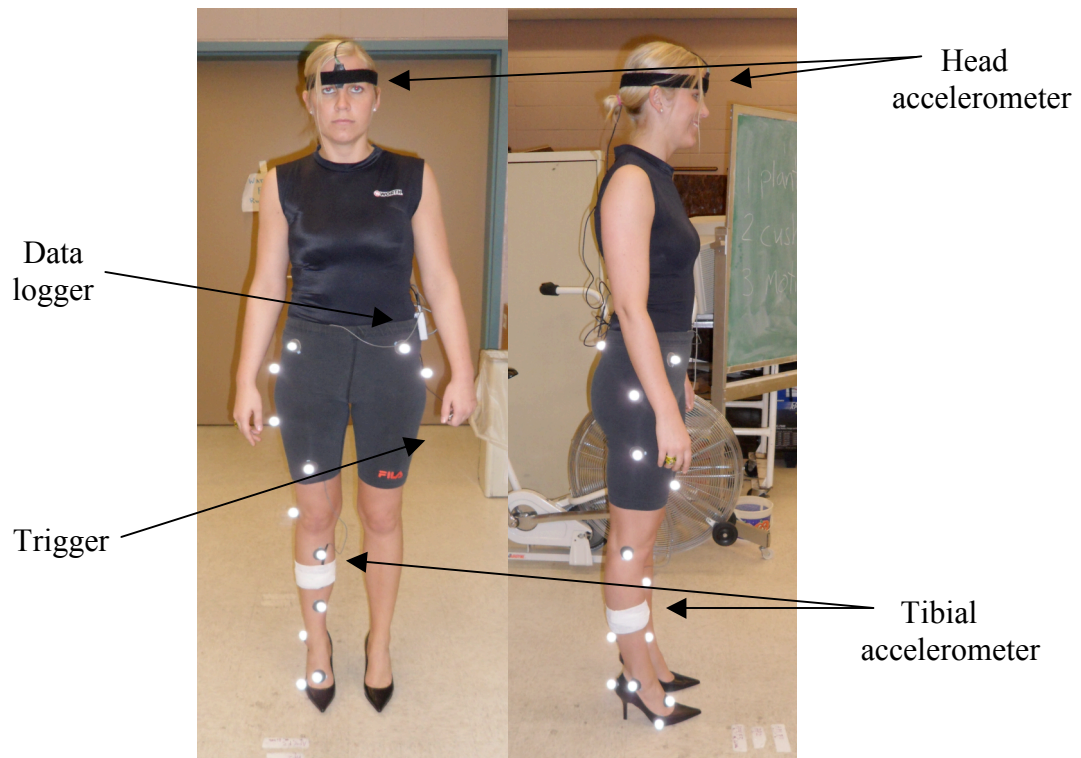


Figure 2. Marker and accelerometer placement.

Subjects were provided with the same shoes worn in session one, and completed heel height conditions in the same order as in session one. Prior to each heel height condition, subjects completed as many practice trials as needed to consistently match the target velocity while contacting a force platform in the center of the walkway with the right foot. Ten acceptable trials for each of the three heel height conditions were completed at the preferred walking velocities measured in test session one. Subjects then completed ten acceptable trials for each of the three heel height conditions at  $1.3 \pm 0.04 \text{ m}\cdot\text{s}^{-1}$ . Acceptable trials were those in which average horizontal velocity of the L5S1 marker during the stance phase on the force platform fell within 3% of the target velocity, the right foot was completely on the

force platform, and there was no visible indication the participant altered step characteristics to impact the force platform.

During each walking trial, marker position and ground reaction force (GRF) data were collected synchronously using an eight-camera Vicon motion system. The motion analysis system was calibrated using a two step process: 1) a rigid reference frame of known coordinates was waved throughout the cameras' view to establish a calibrated volume of space, and 2) the reference frame was used to establish the origin and direction of the Cartesian coordinate system. Marker position and GRF data were sampled at 200 Hz and 1000 Hz, respectively.

Accelerometer and event marker signals were also collected during the walking trials at 1000 Hz using the data logger. The accelerometers were calibrated by attaching both to the missile head of an Exeter impact testing system (Exeter Research, Inc., Exeter, NH) and measuring impacts ranging from 0.46 g to 13.44 g. A 2<sup>nd</sup> order polynomial regression equation was fit to the data in order to convert accelerometer voltages to multiples of the acceleration due to gravity (g). Because accelerometer data were analyzed separately from marker position and ground reaction force data, no attempt was made to fully synchronize accelerometer and marker position data collection. Nevertheless, the event marker was used to mark accelerometer records immediately before and during the completion of each trial for identification of acceleration data associated with each stride cycle collected with the Vicon motion capture system. Immediately prior to initiating a trial, the subject was directed to trigger the event marker consistent with the trial number. The subject also triggered a single event spike approximately at the instant of the left foot contact immediately preceding right foot contact on the force platform.

### Data Analysis

The raw marker position and force platform data were imported into Matlab (The Mathworks, Natick, MA) for data processing. A 4<sup>th</sup> order zero-lag Butterworth digital filter with a low-pass cutoff frequency of 6 Hz was used to smooth raw marker position (Winter, 1990) and force platform data (Bisseling & Hof, 2006). Stride length and stride rate were calculated using the heel marker on the right foot throughout the three-second data collection period. Stride length was calculated as the difference in the position of the heel marker on the right foot along the anterior-posterior axis of the body for consecutive heel strikes of the right foot. Stride rate was computed as the inverse of stride time, which was determined from the time between consecutive heel strikes of the right foot.

Tibial and head accelerometer data were imported into Matlab for data processing. Data were smoothed using a 4<sup>th</sup> order zero-lag Butterworth digital filter with a band-pass filter between 0.5 and 100 Hz. Data were analyzed in the time domain. Peak accelerations at the tibia (PTA) and head (PHA) associated with heel strike were identified and expressed in multiples of the acceleration due to gravity (g) (Figure 3). In addition, shock attenuation (SA) was calculated as the percent decrease of the peak acceleration values (Derrick et al., 1998) as per the following equation:

$$SA = [1 - (PHA/PTA)] \cdot 100 \quad (1)$$

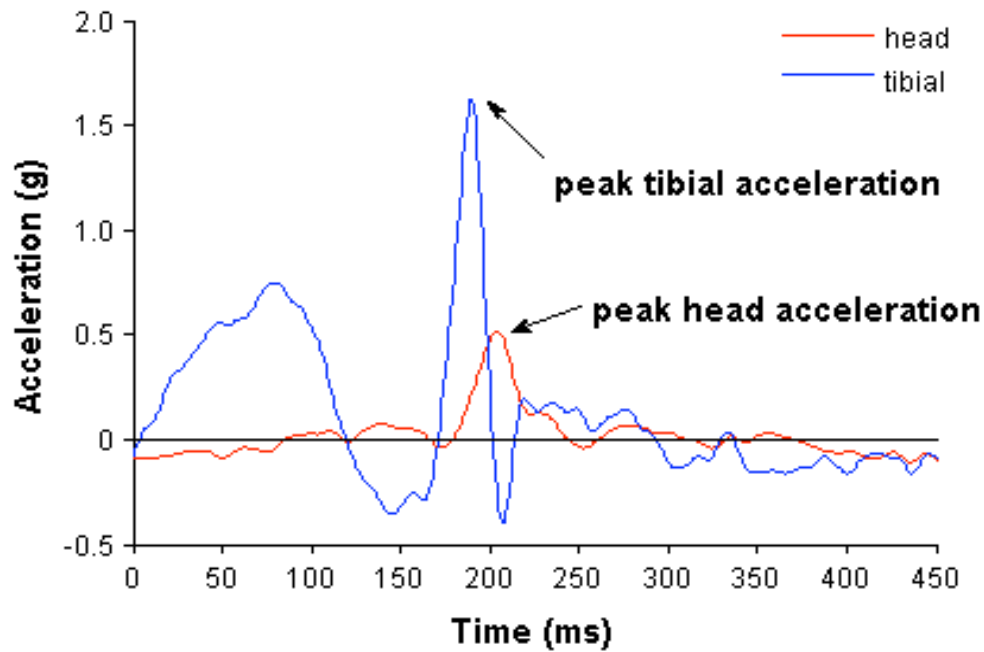


Figure 3. Peak tibial and head acceleration immediately following heel strike.

### Statistical Analysis

A one-way analysis of variance (ANOVA) with repeated measures was used to assess the effect of heel height on PTA, PHA, shock attenuation, and walking kinematics for both preferred and fixed walking velocity conditions. The means of ten trials for each subject for each heel height condition were used in the statistical analysis. The alpha significance level was equal to 0.05. In the event of a significant heel height effect, a post hoc Bonferroni adjustment ( $p = 0.05/3 = 0.017$ ) was used to assess pairwise comparisons.

### **Results**

Subjects reported wearing high heels an average of 2.1 (SD = 1.6) days per week, ranging from once every six months to 5 days per week. The time spent wearing high heels

averaged 5.1 (SD = 1.6) hours per episode. The typical heel height worn by subjects averaged 7.6 cm (SD = 1.5, range: 5.1 – 10.2) cm. Mean length of time of consistently wearing high heels was 7.6 years (SD = 4.4, range: 7 months to 20 years).

Walking velocity was effectively controlled for the fixed speed trials. Subjects were able to closely match the  $1.3 \text{ m}\cdot\text{s}^{-1}$  target speed (Table 1). For the preferred speed conditions, subjects chose a significantly slower preferred walking velocity for the 9 cm heel height compared to the 1 and 5 cm conditions ( $F_{2,13} = 7.34$ ,  $p = .003$ ) (Table 1). Preferred speeds were similar to those previously reported in heel height investigations, which ranged from  $1.23 \text{ m}\cdot\text{s}^{-1}$  (Opilia-Correia, 1990b) to  $1.44 \text{ m}\cdot\text{s}^{-1}$  (de Lateur et al., 1991).

For fixed speed trials, stride length (SL) systematically decreased ( $F_{2,13} = 13.71$ ,  $p < .001$ ) and stride rate (SR) increased ( $F_{2,13} = 7.18$ ,  $p = .003$ ) as heel height increased. The SL and SR values are consistent with those reported in previous research (Opila-Correia, 1990a; Esenyel et al., 2003; Murray et al., 1970). For preferred speed trials, SL was significantly shorter for the 9 cm condition compared to the 1 and 5 cm heel heights ( $F_{2,13} = 13.61$ ,  $p < .001$ ). SR was not affected by heel height changes for preferred walking velocity trials ( $F_{2,13} = 0.09$ ,  $p = .91$ ). The shorter SL for the 9 cm heel height condition accounted for the slower walking velocity for that condition.

Table 1. Mean (SD) kinematic variables in response to heel height at fixed and preferred speeds.

	Fixed Speed			Preferred Speed		
	1 cm	5 cm	9 cm	1 cm	5 cm	9 cm
<b>V (<math>\text{m}\cdot\text{s}^{-1}</math>)<sup>b</sup></b>	1.31(0.01)	1.30(0.01)	1.30(0.01)	1.35(0.10)	1.35(.15)	1.28(.15)
<b>SL (m)<sup>a b</sup></b>	1.40(0.07)	1.37(0.07)	1.34(0.06)	1.40(0.09)	1.40(0.09)	1.33(0.11)
<b>SR (stride<math>\cdot\text{s}^{-1}</math>)<sup>a</sup></b>	0.94(0.05)	0.95(0.06)	0.97(0.04)	0.97(0.02)	0.96(0.02)	0.97(0.02)

V = walking velocity, <sup>a</sup> = significant heel height effect for fixed speed trials, <sup>b</sup> = significant heel height effect for preferred speed trials

Average peak tibial and head acceleration amplitudes were consistent with the values reported in the literature for walking (Voloshin & Loy, 1994; Lafortune, 1991; Voloshin, 1988; Voloshin & Wosk, 1982; Light et al., 1980). Similar heel height effects were observed for fixed speed and preferred speed trials. Heel height significantly affected peak tibial acceleration (PTA) for both fixed and preferred speed conditions (Figure 4; PTA – fixed:  $F_{2,13} = 6.06$ ,  $p = .006$ ; PTA – preferred:  $F_{2,13} = 5.75$ ,  $p = .008$ ). Specifically, PTA was 20-30% higher for the 5 cm heel height condition compared to the 1 cm and 9 cm heel heights, which showed similar PTA amplitudes. As expected, peak head accelerations (PHA) were substantially lower than PTA values (Figure 4). PHA followed similar but more subtle trends as PTA as a function of heel height for both fixed and preferred speed conditions. A statistically significant effect, however, was observed only for the preferred speed trials (PHA – fixed:  $F_{2,13} = 1.74$ ,  $p = .194$ ; PHA – preferred:  $F_{2,13} = 3.87$ ,  $p = .033$ ).

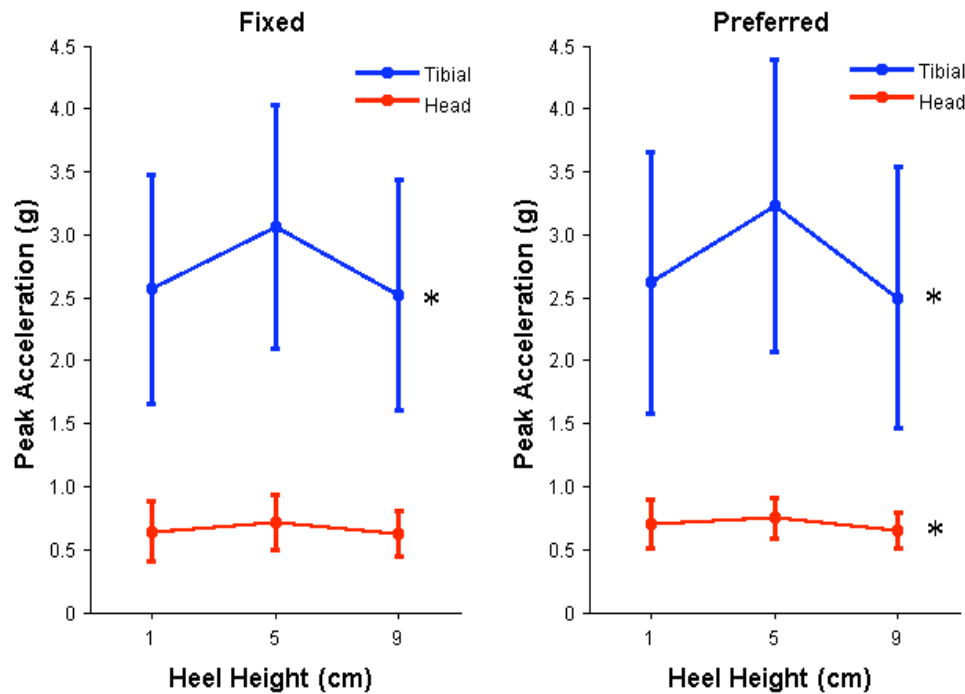


Figure 4. Mean (SD) peak tibial and head accelerations were the highest for the 5 cm condition during fixed and preferred walking velocities. (\* significant effect of heel height).

Shock attenuation across all heel height and walking speed conditions ranged from approximately 70-75% (Table 2). These values are consistent with those reported in the literature, which range from 70 to 80% (Voloshin & Wosk, 1982; Light et al., 1980). A significant heel height effect on shock attenuation was evident for the fixed speed condition ( $F_{2,13} = 3.50$ ,  $p = .044$ ). Shock attenuation was highest for the 5 cm heel height condition. Shock attenuation for preferred speed trials exhibited a similar trend with respect to heel height changes ( $F_{2,13} = 1.67$ ,  $p = .206$ ) but did not reach statistical significance.

Table 2. Mean shock attenuation (SD) in response to heel height for fixed and preferred speeds.

	Fixed			Preferred		
	1 cm	5 cm	9 cm	1 cm	5 cm	9 cm
<b>SA (%)</b> <sup>a</sup>	73.1(8.1)	74.6(7.2)	71.2(10.3)	70.7(10.4)	74.9(7.5)	72.9(9.0)

SA = shock attenuation, <sup>a</sup> = significant heel height effect for fixed speed trials

## Discussion

We hypothesized peak tibial acceleration (PTA) systematically increases as heel height increases, which was not supported by the results. For both fixed and preferred speed conditions, PTA for the 5 cm heel height was significantly higher than PTA for the 1 cm heel height. In contrast to our hypothesis, however, PTA did not continue to increase as heel height increased from 5 cm to 9 cm. Rather, PTA for the 9 cm heel height was significantly lower than that for the 5 cm condition and similar to the average PTA for the 1 cm heel height for both fixed and preferred speed conditions.

The results were partially supported by Voloshin and Loy's (1994) heel height investigation of three subjects. They reported PTA was higher in shoes of approximately 7 cm heel height compared to shoes of approximately 1.4 cm heel height. Although not reported, heel heights of about 4 cm also resulted in higher PTAs than those exhibited in flat shoes, producing a systematic trend for each subject as heel height increased. Our results are comparable to Voloshin and Loy in that an increase in PTA from the 1 to 5 cm condition occurred. Taken together, the results suggest that a moderate increase in heel height may lead to greater impulsive loading due to the rigid structure and limited shock absorbing capacity of the high heel shoe design. However, the 9 cm condition is beyond the heights in



which they investigated, making it difficult to draw conclusions on how higher heel heights affect PTA.

We hypothesized that peak head acceleration (PHA) is unaffected by heel height. Results for the fixed speed condition supported this hypothesis; there were no significant differences in PHA across heel height conditions. A significant heel height effect on PHA was observed for the preferred speed condition, but differences were small (Figure 4). Overall, PHA was 70-75% lower than PTA, with PHA responses being muted by the substantial shock attenuation that occurred between the tibia and head. The results that PHA are quite modestly affected by heel height are supported by previous studies reporting that the body minimizes acceleration at the head in order to facilitate visual and vestibular functions needed for postural and dynamic stability during walking (Menz et al., 2003; Pozzo et al., 1990; Voloshin & Wosk, 1982; Ratcliffe & Holt, 1997). Menz et al. (2003) investigated pelvis and head accelerations during walking on level and irregular surfaces and found that while pelvis accelerations were higher when walking on irregular surfaces, head accelerations showed no change between irregular and level surfaces. Similarly, Pozzo et al. (1990) found that changes in head translation that occurred in walking, running, and hopping tasks were compensated for by changes in angular position of the head in order to stabilize the head. Additionally, the shock attenuation response was consistent with our expectation that shock attenuation increases systematically with PTA as heel height increases so that low PHA is sustained. Our results showed SA was modestly higher when PTA was higher, which was evident in a slight increase at the 5 cm heel height. Also, the considerable shock attenuation of approximately 70 to 75% served to significantly lower PHA and allow for only a modest effect of heel height on PHA.

The unexpected finding that PTA was the highest in the 5 cm condition and similar in the 1 and 9 cm conditions is difficult to explain. PTA can be affected by several variables, including walking velocity and stride length. Faster walking velocities have been shown to be associated with higher PTAs (Voloshin, 2000). Voloshin and Loy (1994) reported a direct relationship between PTA and heel height. Our results for both fixed speed and preferred speed trials for 1 and 5 cm heel height conditions showed this direct relationship. Walking velocities did not differ for 1 and 5 cm heel heights, and therefore, did not confound the heel height effect on PTA. When contrasting the 5 and 9 cm heel height conditions, our results showed an 18 to 23% lower PTA for the 9 cm condition compared to the 5 cm heel height. For the preferred speed trials, the decline in PTA might at first be attributed to the slower average walking speed for the 9 cm condition. However, PTA showed the same response to heel height under fixed speed conditions. These conflicting results indicate the lower PTA for the 9 cm condition cannot be attributed to walking velocity changes.

The isolated effect of stride length on PTA has not been investigated in walking. Derrick et al. (1998) demonstrated PTA and stride length are directly related to one another during running. Shorter stride lengths reduce the vertical velocity of the foot at foot strike, contributing to lower PTA. Assuming for the moment that these trends noted for running also apply to walking, the decreases in stride length observed as heel height increased for fixed velocity and preferred velocity trials would have a tendency to reduce PTA systematically. However, our results showed there was no systematic relationship between stride length and PTA. Consequently, our PTA profiles as a function of heel height cannot be explained by stride length changes that also occurred as heel height was manipulated.

It is evident that PTA response cannot be explained by a single variable, and is most likely the result of a combination of gait alterations. Voloshin and Loy (1994) reported, “the subject’s walking style and gait velocity apparently had the most effect on the magnitudes of the heel-strike generated shock waves,” although they did not mention any specific variables relating to “walking style.” It may be that a threshold of heel height exists, in which beyond a certain height, individuals make adaptations in walking, whether aimed at or not, reduce the PTA. The exact combination of kinematic alterations that contributed to the lower PTA at 9 cm condition in this study remains unclear.

In order to better understand the relationship between heel height and PTA, further research is necessary. Possibly controlling for walking velocity and stride length through the use of a treadmill to study the effects of heel height may help to better control for other influences on PTA. Additionally, the use of more heel heights, including heights that are between those used in this study as part of a systematic manipulation may help to provide a more comprehensive view of the relationship between heel height and PTA. Lastly, other variables can be assessed in order to determine any contributions to PTA. Velocity of the foot and ankle immediately before heel strike during walking have been shown to be positively related to PTA (Jefferson, Collins, Whittle, Radin & O’Conner, 1990; Radin, Yang, Riegger, Kish, & O’Conner, 1991), making it worthwhile to investigate these variables with increasing heel height. Although not specifically investigated in walking, the stiffness of the lower limb with respect to muscle contribution has also been suggested to influence PTA. Flynn, Holmes, and Andrews (2004) reported a reduction in PTA at impact after fatiguing the tibialis anterior and gastrocnemius muscles, which was suggested to be due to a decreased force producing capacity of the muscles, resulting in the muscles being less stiff

than in a pre-fatigue state. Therefore, investigating muscle contribution during walking may be beneficial in understanding how the body adapts to heel height during walking and if there is any influence on PTA.

Several limitations of this study are worth noting. First, a skin-mounted accelerometer was used to estimate skeletal accelerations. The method of using a low-mass accelerometer that is firmly mounted to tibia using elastic bands has been employed previously (Derrick et al., 1998; Mercer, Devita, Derrick, & Bates, 2003). Ziegert & Lewis (1979) and Light et al. (1980) found the effect of soft tissue between the accelerometer and the bone was negligible for a low-mass accelerometer securely attached to the tibia. In addition, subjects were provided with shoes instead of wearing their own. Although this enhances the internal validity of the study by controlling footwear structure, it also could have potentially created changes in gait characteristics due to subjects not being accustomed to wearing the shoes. Subjects were given adequate time, however, to become familiarized to each shoe condition prior to data collection in order to minimize this response. Lastly, the sample reflected considerable variability in the frequency of wearing high heels. The frequency of wearing high heels may affect walking kinematics and kinetics, particularly more variable gait characteristics. However, a cursory comparison of the gait kinematics and kinetics of the most frequent high heel wearers to those who wear high heels least frequently revealed no obvious differences in response to heel height. In support of this observation, Ebbeling et al. (1994) reported no significant differences in the kinematic and kinetic gait parameters between experienced (3 times/ week; 8 hours/day) and inexperienced (less than 3 times/week; 8 hours/day) wearers of high heels.

In conclusion, PTA was highest for the middle heel height condition (5 cm), suggesting this heel height had the greatest potential for deleterious effects on the body's natural shock absorbers, thereby contributing to greater joint degeneration and the development of knee OA. Additionally, the slightly higher shock for the middle heel implies greater action of the body's natural shock absorbers, which may contribute to their progressive weakening. Further investigation of walking kinematics and kinetics in response to heel height is necessary in order to identify possible adaptations that affect repetitive impulsive loading at heel strike.

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## CHAPTER 4. GENERAL CONCLUSIONS

Wearing high heels alters walking kinematics and kinetics, and it is evident that these alterations can lead to adverse loading on the body. In the frontal plane, an increase in heel height resulted in higher external knee adduction moments, which are indicative of greater medial loading at the knee. In addition, kinetic changes at the ankle may also contribute to greater medial loading at the knee. This is of great importance, as knee osteoarthritis (OA) occurs most often in the medial compartment of the knee, and larger medial loads can contribute to joint degeneration. Therefore, wearing high heels may put individuals at greater risk for developing knee OA. Moreover, it is evident that higher heels lead to greater medial loading, suggesting wearing higher heel heights increase the risk of developing knee OA.

Not only does wearing high heels influence the lower extremity loading in the frontal plane, but it's also evident that heel height affects repetitive impulsive loading during walking. Moderate heel heights increase transient accelerations at heel strike, which can contribute to joint degeneration and knee OA. However, it is evident that at higher heel heights, the body makes gait adaptations that result in reduced loading at heel strike. However, it is not clear how the body adapts kinematically and kinetically in order to reduce impulsive loading at higher heel heights. The lower impulsive loading with higher heels compared to moderate heels does not imply that higher heel heights are less detrimental to the body. A variety of kinematic and kinetic alterations occur at higher heel heights, such as higher external adduction moments, which can contribute to adverse loading in other ways.



Future research should focus on the investigating the interplay between heel height, kinematic changes, and impact accelerations in order to better understand how adaptations in walking contribute to repetitive impulsive loading. Additionally, the inclusion of more heel height conditions may help to better define the relationship between heel height, impact acceleration, and medial loading at the knee as well as other kinematic and kinetic variables. Lastly, investigating the effects of time of exposure to high heels may be beneficial in understanding how the body adapts to higher heel heights over a period of time.

In conclusion, wearing high heels may put individuals at greater risk for developing knee OA through the repetitive dynamic loading that occurs during walking. This is of great importance due to many women wearing high heels and the increasing prevalence of knee OA.

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## APPENDIX A:

**Mean, Standard Deviations, and Analysis of Variance Statistical Results**

(a = significantly different from 1 cm, b = significantly different from 5 cm)

Primary Variables

Variable		Fixed Speed			Preferred Speed		
		1 cm	5 cm	9 cm	1 cm	5 cm	9 cm
Velocity ( $\text{m}\cdot\text{s}^{-1}$ )	Mean	1.31	1.30	1.30	1.35	1.35	1.28 <sup>a b</sup>
	SD	0.01	0.01	0.01	0.10	0.15	0.15
	F (p)	1.94 (0.163)			7.34 (0.003)		
Stride length (m)	Mean	1.40	1.37 <sup>a</sup>	1.34 <sup>a</sup>	1.40	1.40	1.33 <sup>a b</sup>
	SD	0.07	0.07	0.06	0.09	0.09	0.11
	F (p)	13.71 (<0.001)			13.61 (<0.001)		
Stride rate ( $\text{stride}\cdot\text{s}^{-1}$ )	Mean	0.94	0.95	0.97 <sup>a</sup>	0.97	0.96	0.97
	SD	0.05	0.06	0.04	0.07	0.08	0.07
	F (p)	7.19 (0.003)			0.093 (0.911)		

Frontal Plane Kinetics Primary Variables

Variable		Fixed Speed			Preferred Speed		
		1 cm	5 cm	9 cm	1 cm	5 cm	9 cm
Peak internal hip abduction moment early stance (Nm/kg)	Mean	-0.78	-0.79 <sup>a</sup>	-0.86 <sup>b</sup>	-0.79	-0.80	-0.84
	SD	0.07	0.08	0.09	0.10	0.09	0.08
	F (p)	15.99 (<0.001)			3.30 (0.052)		
Peak internal hip abduction moment late stance (Nm/kg)	Mean	-0.68	-0.69	-0.67	-0.68	-0.68	-0.68
	SD	0.07	0.06	0.06	0.07	0.07	0.07
	F (p)	2.34 (0.115)			0.188 (0.83)		

Peak external knee adduction moment early stance (Nm/kg)	Mean	0.46	0.48	0.55 <sup>b</sup>	0.47	0.49	0.53 <sup>b</sup>
	SD	0.11	0.07	0.08	0.12	0.07	0.08
	F (p)	8.71 (0.001)			3.96 (0.03)		
Peak external knee adduction moment late stance (Nm/kg)	Mean	0.38	0.44 <sup>a</sup>	0.49 <sup>a</sup>	0.37	0.45 <sup>a</sup>	0.48 <sup>a</sup>
	SD	0.06	0.09	0.10	0.06	0.08	0.10
	F (p)	20.77 (<0.001)			26.80 (<0.001)		
Peak internal ankle eversion moment (Nm/kg)	Mean	-0.14	-0.23 <sup>a</sup>	-0.29 <sup>a</sup>	-0.13	-0.22	-0.27 <sup>a</sup>
	SD	0.08	0.12	0.10	0.09	0.13	0.10
	F (p)	16.00 (<0.001)			15.79 (<0.001)		

### Frontal Plane Kinetics Secondary Variables

Variable		Fixed Speed			Preferred Speed		
		1 cm	5 cm	9 cm	1 cm	5 cm	9 cm
Peak sagittal plane hip power at push-off (W/kg)	Mean	1.35	1.28	1.33	1.28	1.23	1.28
	SD	0.91	0.51	0.40	0.49	0.34	0.44
	F (p)	0.10 (0.903)			0.21 (0.809)		
Peak sagittal plane knee power at push-off (W/kg)	Mean	-1.51	-1.44	-1.43	-1.70	-1.57	-1.45
	SD	0.38	0.32	0.28	0.37	0.41	0.40
	F (p)	1.01 (0.38)			4.90 (0.015)		
Peak sagittal plane ankle power at push-off (W/kg)	Mean	3.21	2.93 <sup>a</sup>	2.04 <sup>a b</sup>	3.32	3.10	2.05 <sup>a b</sup>
	SD	0.58	0.45	0.44	0.62	0.63	0.53
	F (p)	137.20 (<0.001)			98.13 (<0.001)		
Peak ankle eversion angle (°)	Mean	-5.08	-4.67	-3.86	-4.97	-4.47	-3.87
	SD	2.17	2.48	2.89	1.92	2.32	2.84
	F (p)	1.60 (0.22)			1.32 (0.282)		
Peak ankle inversion angle (°)	Mean	4.75	2.90	2.72 <sup>a</sup>	5.05	2.98 <sup>a</sup>	2.37 <sup>a</sup>
	SD	2.69	2.71	2.84	2.74	2.46	1.90
	F (p)	5.79 (0.008)			13.11 (<0.001)		

### Impact Acceleration Primary Variables

Variable		Fixed Speed			Preferred Speed		
		1 cm	5 cm	9 cm	1 cm	5 cm	9 cm
Peak tibial acceleration (g)	Mean	2.57	3.06 <sup>a</sup>	2.52 <sup>b</sup>	2.62	3.23 <sup>a</sup>	2.50 <sup>b</sup>
	SD	0.91	0.97	0.92	1.04	1.16	1.04
	F (p)	6.06 (0.006)			5.75 (0.008)		
Peak head acceleration (g)	Mean	0.64	0.72	0.65	0.70	0.75	0.62
	SD	0.19	0.16	0.14	0.24	0.22	0.18
	F (p)	1.74 (0.194)			3.87 (0.033)		
Peak shock attenuation (%)	Mean	73.1	74.6	71.2 <sup>b</sup>	70.7	74.9	72.9
	SD	8.4	7.2	10.3	10.4	7.5	9.0
	F (p)	3.50 (0.044)			1.67 (0.206)		

### Impact Acceleration Secondary Variables

Variable		Fixed Speed			Preferred Speed		
		1 cm	5 cm	9 cm	1 cm	5 cm	9 cm
Peak vertical impact force (BW)	Mean	0.62	0.51	0.47 <sup>a</sup>	0.66	0.57	0.46 <sup>a b</sup>
	SD	0.17	0.10	0.06	0.17	0.11	0.09
	F (p)	9.96 (0.001)			15.95 (<0.001)		
Knee angle at contact (°)	Mean	3.70	5.70 <sup>a</sup>	7.02 <sup>a</sup>	4.56	6.44	7.81 <sup>a</sup>
	SD	5.90	5.55	6.33	5.38	5.22	6.08
	F (p)	13.78 (<0.001)			9.62 (0.001)		